THE BIOMECHANICAL EFFECT OF TRUNK INCLINATION ON JOINT MOMENTS AND MUSCLE ACTIVATION IN PEOPLE WITH KNEE OSTEOARTHRITIS

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Abbreviations of the muscles

LG	**Lateral gastrocnemius**
MG	**Medial gastrocnemius**
VL	**Vastus lateralis**
VM	**Vastus medialis**
BF	**Biceps femoris**
ST	**Semitendinosus**
GX	**Gluteal maximus muscle**

Abstract

Knee osteoarthritis is a progressive disease, which is associated with pain, stiffness and disability. Previous research has demonstrated that the progression of knee OA is influenced by numerous biomechanical factors, which will affect the loading on the knee joint. Cross sectional studies have shown altered knee OA gait to be characterised by altered joint moments, increased muscle activity and increased muscular co-contraction. Furthermore, modelling studies have shown that the alterations in muscle patterns that are characteristic of knee OA lead to elevated joint loads, while longitudinal studies have demonstrated the progression of knee OA to be related to these altered muscular responses.

In this thesis, I examine a new model that has the potential to explain some of the previous observed differences in joint moments and muscle activity characteristic of knee OA gait. Specifically, this thesis investigates the effect of increased sagittal plane inclination of the trunk (forward trunk lean) and the effect this could have on hip, knee and ankle moments and muscle activation patterns. Three studies were conducted. The first sought to characterise differences in sagittal trunk inclination during walking, between people with knee OA and healthy control participants. This study demonstrated that people with knee OA walk with approximately 3° more trunk flexion than healthy control subjects. In the second study, I examined the relationship between trunk inclination and a range of different biomechanical parameters, including hip extensor moments, hamstring activation levels and hamstring-quadriceps co-contraction. These results of this study showed weak to moderate correlations between hip extensor moment and the magnitude of hamstring activation during the period 15-25% of the stance phase of the gait cycle. In the final study, I explored the effect of a three-curve rocker shoe on trunk inclination and a range of other biomechanical variables. These data showed that, although there was significant reduction in trunk inclination with the rocker shoe, there was no

corresponding decrease in moments or muscle activations over the period 15-25% of stance. Taken together, the results of this thesis show that people with knee OA walk with an increased trunk lean and that this may explain some of the previously observed differences in hip moments and muscle patterns. However, further work is required to establish if interventions that could reduce trunk lean (such as a rocker shoe) could lead to long-term clinical benefits for people with knee OA.

Chapter 1 - Introduction

1.1 Introduction

What is knee osteoarthritis?

Osteoarthritis (OA) is a type of joint disease which is characterised by the breakdown of cartilage and underlying bone (Brooks, 2002, Reginster, 2002), and which can also affect ligamentous structures. In the active phase of OA, physiological processes lead to the destruction of cartilage, bone thickening in the subchondral area and new bone formation (Peat et al., 2001) (see Figure 1-1). As OA progresses, there is a continued loss of articular cartilage and osteophytes (small bony projections) gradually form at the margins of the joint (Altman et al., 1986; Sangha, 2000). These physiological changes are normally accompanied by clinical pain, stiffness and swelling around the joint (Kean et al., 2004). As the most common joint disease, OA is among the most frequent and symptomatic health problems for middle aged and older people (Buckwalter et al., 2004). The development of osteoarthritis is dependent on many intrinsic factors, such as age, gender and genetic predisposition (Kennish et al., 2014), but also on environmental factors, such as previous injuries to the joint and abnormal biomechanical loading (Andriacchi, 1994).

Figure 1-1. Photographs of healthy knee joint (left) and knee osteoarthritic (right) knee joint specimens (Peters et al., 2018)

The common joints affected with OA are the hand, knee and hip (Hunter and Eckstein, 2009). Interestingly, the knee joint is the most affected, with a prevalence twice that of OA of the hand and hip joints (Oliveria et al., 1995). Clinically, knee OA is characterized by clinical symptoms including joint swelling, pain, tenderness, movement limitation in the joint, crepitus in the joint and stiffness early in the morning (Bijlsma and Knahr, 2007, Buckwalter et al., 2004). Knee osteoarthritis can influence daily activities, work and emotional status and lead to disability in the long term. It therefore places a large emotional and financial burden on the individual and also on society as a whole.

1.2 Incidence of knee osteoarthritis

Over 100 million people across the world suffer from OA, as revealed by global statistics, and the disease is one of the most common reasons for disability (Hinman et al., 2010, Heiden et al., 2009a). Yu et al. (2015) have recently reported that in the UK, 1% of adults are newly diagnosed with OA each year. However, this percentage increases to 3% in those aged between 75 and 85 years old (Yu et al., 2015). Moreover, this study used a run-in period and stratification by age group, which revealed a rising OA incidence in adults aged between 35 and 44 years old (Yu et al., 2015). In England, 9 people in each 1000 face diagnosis with osteoarthritis yearly (Yu et al., 2015) and in the United States of America, while an estimated 21 million adults had OA in 1995, unfortunately, this number had risen to approximately 27 million by 2008 (Lawrence et al., 2008). Another recent study (Radha and Gangadhar, 2015) gives the prediction that by 2025, India will be home to the largest number of arthritis sufferers of any country in the world, at an expected 60 million sufferers. Traditionally, OA incidence is assumed to increase among older individuals and to be more prevalent in females than males (Lawrence et al., 2008). Similarly, Felson et al. (1995) report that gender plays a role in OA incidence (Felson et al., 1995).

The knees and hips are the most frequent areas in which osteoarthritis is seen. Moreover, the most common occurrence of osteoarthritis is in the knee joint, and this joint is a primary location for pain and disability resulting from OA, affecting between 3 and 4 in ten of 60-year-olds (Felson, 1990, Lawrence et al., 1998). Among those over 30 years of age, approximately six per cent have symptoms of OA in the knee, with this rising to eleven per cent among those over 65 (Guccione et al., 1990). At the same time, between 20 and 28 per cent of adults over 40 in the UK exhibit pain symptoms in this joint, and in half of these cases, it is predicted that these individuals will go on to develop OA of the knee (Peat et al., 2001).

Knee OA in women is the fourth most common reason for disability, while it is the eighth in males, based on World Health Organization (WHO) reports (Vad et al., 2002). Furthermore, the chance of a woman developing OA in the knee is substantially greater in comparison with males (Felson et al., 1995), and women after menopausal age, aged 55 years and over, are more likely to have increased severity of knee OA (Srikanth et al., 2005). Further, knee osteoarthritis incidence increases with age, so that approximately 11% of females older than 60 are knee OA symptomatic (Creamer et al., 2000). Whereas radiographic knee osteoarthritis had a prevalence of 19.2% for the Framingham study in the >45 age group, it was 27.8% in Johnston County, and in those aged 60 or over, about 37.7% in the NHANES III study (Dillon et al., 2006). On the other hand, Knee OA symptoms in adults aged 26 or over reached nearly 5% in Framingham, and in people aged 45 or over it reached 16.7% in Johnston, while it was 12.1% in the NHANES III study in those aged 60 and over (Dillon et al., 2006).

Across the UK population, 20% to 28% of those aged 40 or over have knee pain and 50% develop knee OA (Peat et al., 2001). The knee OA rate of incidence based on population studies in the US is similar to that in Europe. Those studies show that severe radiological changes affect adults aged between 25 and 34 at a rate of 1%, and that this increases to about half of individuals in the >75 age group (Litwic et al., 2013). In China, studies using a similar definition and methods to the Framingham study reveal that the incidence of knee OA (bilateral compartment disease) was higher by 2 to 3 times in the Chinese cohort in comparison to rates estimated by the Framingham research among OA sufferers (Kang et al., 2009). In the Chinese Asia-Pacific region, 7.5 % of the population were found to be suffering from knee OA (Wigley et al., 1994), while the percentage was almost 6% in rural India (Chopra et al., 1997). In urban India, this proportion was 22% to 28%, while it was 25% in the rural population of north Pakistan (Farooqi and Gibson, 1998). In other countries such as Bangladesh, there are about 10 people affected with OA for each 100 population (Haq et al., 2005).

1.3 The economic cost of knee osteoarthritis

The economic costs of osteoarthritis are substantial, and can be divided into two types; indirect and direct costs. Direct costs include surgery and pharmacological/non-pharmacological treatments, as well as use of hospital resources and management of complications arising from osteoarthritis treatment. Indirect costs appear as a loss of personal working time, low productivity due to premature retirement, pain, care-giver time, disability compensation/benefits and mortality (Chen et al., 2012). In particular however, knee OA treatment is a major burden to health care. In the UK in 2006, the Royal College of General Practitioners estimated that more than one million adults consulted their GP every year with osteoarthritis symptoms (Royal College of General Practitioners, 2006). In 2007, another study presented the finding that osteoarthritis consultations accounted for 15% of all musculoskeletal consultations in patients aged 45 and older, increasing to 25% in those aged 75 and over (Chen et al., 2012). The consultation cost is estimated at £36 for a 12-minute consultation visit (Curtis, 2008). To improve function and relieve knee pain in knee OA, total knee arthroplasty (TKA) is a commonly performed surgical procedure. Every year, more than 650, 000 TKAs are performed in the United States (Kolisek et al., 2007). The total cost of osteoarthritis in the United States of America, Canada, United Kingdom, France, and Australia is estimated at between 1 and 2.5% of the gross national product (GNP) for those countries (March and Bachmeier, 1997).

Osteoarthritis has a substantial negative impact on the UK economy. It has been estimated that the total cost forms the equivalent of 1% of GNP every year and that the disease leads to an estimated 36 million working days lost. This is because of the large number of people with osteoarthritis, the impact on quality of life, ability to work, and the need for health, social care and welfare benefits (Chen et al., 2012). Furthermore, knee OA may draw a higher cost in terms of social functions and associated disability than different joints affected with OA, and osteoarthritis imposes a burden on health care and the general economy, with one quarter of United Kingdom citizens of 65 or over showing changes in the knee associated with OA (Jinks et al., 2004, Bijlsma and Knahr, 2007). So, in view of the increasing health burden and prevalence of OA, there is an urgent need to understand the causes of knee OA in order to find preventative and effective therapies and reduce risk factors for both the incidence and progression of knee OA.

1.4 Statement of the problem

OA is the most common joint disease worldwide with the knee joint being the most affected joint (Felson, 1990, Lawrence et al., 1998). Furthermore, knee OA is clearly known to lead to disability, mainly in older people, and indirectly leads to pain. This disease affects the economic, physical, social and health spheres, impacting upon body functions, causing limited mobility, stiffness and decreasing activity undertaken on a day-to-day basis. In general, this burden has impact at personal, national and world scales. However, importantly, progression of knee OA is known to be strongly influenced by biomechanical factors which will affect loading patterns of the knee during walking (Guilak, 2011).

Previous research has looked at a range of biomechanical factors that have the potential to influence knee joint loading(Sritharan et al., 2016a). For example, many studies have investigated differences in joint moments (e.g. Baliunas et al., 2002b; Mundermann et al., 2005; Kaufman et al., 2001; Astephen et al., 2008; Sritharan et al., 2016) on the basis that an increased moment will increase the loading at the joint surface (de David et al., 2015). In line with this idea, research has examined the knee adduction (or frontal plane) moment as this is related to the magnitude of loading on the medial compartment. Other studies have looked at the sagittal plane moment at the knee (Liu et al., 2014; Preece et al., 2016) as an increased sagittal moment could also lead to increased stress on the articular surface. There is now a large body of research showing that people with knee OA walk with increased co-contraction (simultaneous activity of the agonist and antagonist muscles) (Sirin and Patla, 1987, Childs et al., 2004, Lewek et al., 2004, Hubley-Kozey et al., 2006). Through modelling studies, this increased muscle activity has been shown to increase the loads at the articular surface (Brandon et al., 2014, Sritharan et al., 2016a) and possibly accelerate disease progression. However, to date there is no widely accepted theory to explain why knee muscle activity is elevated in people with knee OA.

In this thesis, I put forward a new model to explain both the alterations in moments and muscle activity which have been observed in in people knee. The idea is developed from an understanding of how subtle alterations in sagittal plane trunk inclination, adopted during walking could alter the direction of the ground reaction force vector and therefore the moments and muscle activation patterns at the hip knee and ankle. Following a detailed literature review and methodology section (Chapters 2&3); I first explore differences in sagittal plane trunk inclination between healthy people and individuals affected by knee OA (Chapter 4). In the following chapter (Chapter 5), I then investigate possible associations between trunk inclination and moments/muscle activations. This research is novel because, although there has been a lot of previous work investigating the links between trunk position in the frontal plane and knee adduction moments (Creaby et al., 2012, Bechard et al., 2012, Hunt et al., 2008), there has been very little investigation into the characteristics, and effects, of sagittal plane trunk alignment in people with knee OA.

In the final chapter (Chapter 6), I explore the effect of a footwear intervention that has the potential to alter sagittal plane alignment of the trunk. Footwear interventions are relatively easy to implement and may therefore prove to be an effective conservative management approach for people with knee OA. However, although there has been a considerable amount of previous research investigating the biomechanical effects of footwear interventions in knee OA, most studies have sought to investigate footwear-induced changes in frontal plane moments (Shakoor et al., 2008, Jones et al., 2013, Jones et al., 2014). In contrast, the final study in this thesis focuses on the sagittal plane and explores whether a three-curved rocker could be used to improve sagittal plane trunk alignment and produce corresponding changes in joint moments and muscle activation patterns.

Chapter 2 - Literature review.

2.1 Definition, clinical characteristics and diagnosis of knee osteoarthritis

Knee OA is characterised by an active disease process involving cartilage destruction, subchondral bone thickening, and new bone formation (Peat et al., 2001). Clinically, diagnosis of knee OA depends upon a symptom set which is largely subjectively reported, such as swollen areas and pain. Alongside these indicators, objectively recorded symptoms from physical investigation include assessment of deformation or stiffness in the knee, as well as radiographically recorded data to supplement other indicators. According to the American Academy of Orthopaedic Surgeons (2004), osteoarthritis may be classified into two types; primary (idiopathic) or secondary (caused by metabolic, anatomical, traumatic or inflammatory conditions). The primary form involves the hip, knee, hand, spine and other joints. Primary knee osteoarthritis is described as occurring through degenerative processes in the articular cartilage without an identifiable abnormal condition underpinning those processes (Peters et al., 2018). However, secondary knee osteoarthritis frequently occurs due to traumatic impacts or repeated motions, including those linked to particular jobs. Secondary osteoarthritis of the knee can also come from other disease or congenital abnormalities.

There are many clinical features that are used to identify the presence of knee OA on clinical examination. These symptoms include pain, swelling, stiffness mainly in the morning and after sitting for a long time, decreased mobility during activities for daily living such as walking, and a crackling sound in the knee during walking (Jackson et al., 2003).

2.1.1 Pain in Knee Osteoarthritis

The most challenging of the symptoms of knee osteoarthritis tends to be pain (Peter et al., 2011). Peter et al. (2011) report that in the early stages of knee OA, pain occurs and worsens in the joint when the patient starts to walk, and commonly increases during daily activity and after a long duration bearing weight on the knee joint, such as when standing or walking. This pain can however still be felt during resting and at night. More than 50% of those over 65 state that their OA causes them pain (Parmelee et al., 2007). The location of this pain is usually in the knee joint or in the area around the joint, and is sometimes located in the upper leg or above the knee (Peter et al., 2011). Patients with knee OA usually avoid moving their painful joints and so function becomes impaired (Dandy and Edwards, 2009). This action leads to weakening in the knee muscle: mainly the extension muscle. It is reported that weakness in the quadriceps, unstable joints, pain and restriction of function are the symptoms which OA sufferers mainly report (Hurley et al., 1997). The location of knee OA is sometimes in the patellofemoral joint, but more often in the medial tibiofemoral compartment and occasionally in the lateral tibiofemoral compartment (Felson, 2006).

2.1.2 Stiffness

Knee stiffness is usually reported by people who are suffering from knee osteoarthritis, and forms a single criterion from the list of six which are employed to diagnose osteoarthritis of the knee (Gignac et al., 2006). More than 50% of people over the age of 65 report some pain and stiffness due to osteoarthritis (Parmelee et al., 2007). It is reported that morning stiffness is associated with osteoarthritis, however, this usually lasts less than 30 minutes (Sorensen et al., 2014). An increase in self-reported knee stiffness is correlated with a significantly higher risk of increasing incidence of osteophytes (Mazzuca et al., 2007) and also in the progression/growth of osteophytes over time (Mazzuca et al., 2006). Furthermore, self-efficacy for physical tasks in knee OA is related to the sensation of knee stiffness (Maly et al., 2006). Therefore, knee stiffness is an important symptom associated with knee OA and warrants evaluation (Dixon et al., 2010).

2.1.3 Functional Ability and Knee Osteoarthritis

Ability in daily functional activity is significantly impaired with knee osteoarthritis and this can impact on gait (Bejek et al., 2006). Some degree of physical activity limitations are reported in 20–80 percent of osteoarthritis patients and this has been shown to be an independent risk factor for functional decline (Parmelee et al., 2007). Statistical differences between patients with knee OA and healthy control subjects have been observed in walking characteristics, such as cadence, step length, knee & hip joint motions and also motion of the pelvis (Bejek et al., 2006). These observations support the idea that knee OA is associated with considerable changes in overall movement and this is likely to be linked to functional decline (White et al., 2014).

2.1.4 Level of severity of knee OA diagnosed by Kellgren-Lawrence radiographic classification.

Clinically, diagnosis of knee OA depends on subjective symptoms of the knee, including pain or swelling. However, objective physical examination of knee stiffness or deformity along with a number of supplementary radiographic findings may also be used. Radiographic investigations observe abnormalities in osteophytes, narrowing of the joint space, subchondral sclerosis and subchondral cysts as signs of knee OA (Peter et al., 2011). In addition, radiographic definitions are based on the Kellgren-Lawrence radiographic classification, which grades the extent of radiographic osteoarthritis from 0 to 4, based on the presence and severity of individual radiographic features such as osteophytes and joint space narrowing (Chaganti and Lane, 2011). Bing et al. (2011) suggest that it is important to evaluate the progression of the disease and understand it through the radiographic features of the OA knee: mainly, joint space narrowing. In light of this, X-rays and MRIs are still most commonly used in OA to evaluate pathological progression.

The Kellgren-Lawrence (KL) scale has long been considered a gold standard for evaluating knee OA progression and severity on the basis of X-ray results (Altman et al., 1986). Severity has been classified into five grades according to the KL scale, graded between 0 and 4, with grade 0 indicating no features (normal), and grade 4 showing the highest severity, with large osteophytes, marked narrowing of joint space, severe sclerosis and definite deformity of bone contour (Kellgren and Lawrence, 1957, Miyazaki et al., 2002, Sharma et al., 1998).

2.1.5 ACR criteria for knee OA

The American College of Rheumatology (ACR) developed the clinical classification criteria of knee OA. This remains a popular method of classifying knee osteoarthritis, recommended for epidemiological and clinical studies (Brooks and Hochberg, 2001) and the practice of primary care (Jackson et al., 2003). ACR criteria have a specificity of 93% and sensitivity of 94% (Heidari, 2011). They consist of five criteria which are: (1) knee pain for most days of the preceding month; (2) crepitus on active joint motion; (3) morning stiffness of at least 30 min in duration; (4) age > 38 years; and (5) bony enlargement of the knee on examination. If items 1, 2, 3, 4, or 1, 2, 5 or 1, 4, 5 are present, respondents are diagnosed with clinical osteoarthritis (Mathers et al., 2000). Wu et al. (2005) conclude that the ACR clinical criteria detect OA patients with cartilage damage prior to any radiographic change, whilst the ACR clinical and radiographic classification criteria detect OA patients with severe cartilage damage. The ACR criteria associates well with articular cartilage damage in patients with knee OA (Wu et al., 2005).

2.2 Risk factors for incidence of knee OA

There is a range of different factors that have the potential to influence OA in the knee joint. These factors encompass systematic factors and biomechanical factors affecting progression of the disease. The systematic factors (age, gender, menopause, genetics etc.) are those that increase susceptibility to progression of knee OA disease. Those factors interact with environmental/mechanical factors (obesity, joint injuries, severity of the joint, muscle weakness etc.) to influence articular structure/damage, progression of the disease and ultimately joint breakdown (Figure 2-1). Thus, mechanical factors play a role in determining OA severity in the knee joint (van Raaij et al., 2010, Cooper et al., 2000, Felson et al., 2000).

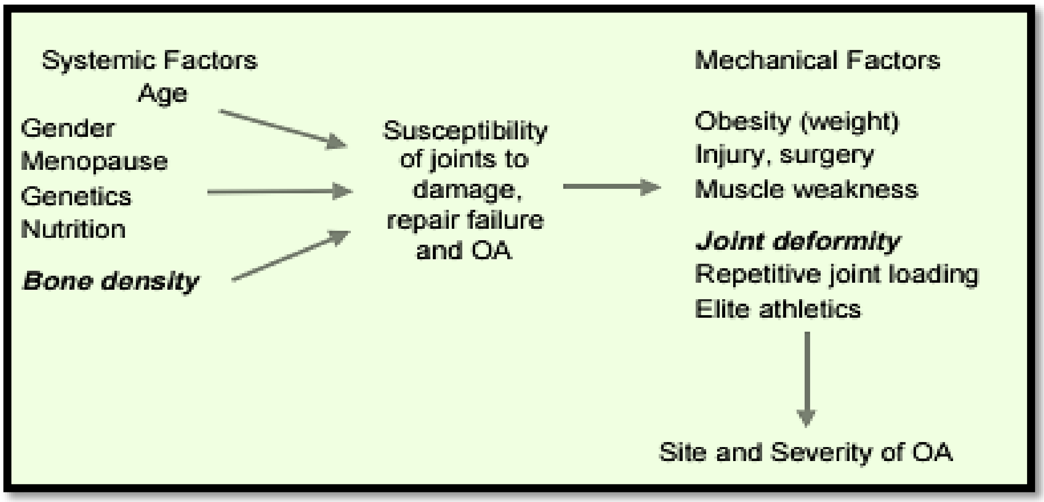


Figure 2-1. Adapted from Dieppe, 1995

Two different approaches have been used to study the risk factors associated with specific diseases, such as knee OA: a case-control and a cohort study. With a case-control study, patients who have the disease or outcome of interest (cases) are compared with patients who do not have the disease or outcome (controls). The investigator then looks back retrospectively to compare how frequently exposure to a risk factor is present in each group, to determine the relationship between the risk factor and the disease. The second method is a cohort study, defined as a design where one or more samples (called cohorts) are followed prospectively and subsequent status evaluations with respect to a disease or outcome are conducted to determine which initial participants' exposure characteristics (risk factors) are associated with it. As the study is conducted, the outcome from participants in each cohort is measured and relationships with specific characteristics determined. In the paragraphs below, the different risk factors, which have been obtained either through case-control or cohort studies and are divided up into systemic and environmental/mechanical factors:

2.2.1 Systemic Factors:

A. *Age*:

The prevalence and incidence of knee OA is sharply increased with age (Felson et al., 1995, Lawrence et al., 2008). Around 50 % of people aged fifty and over indicate having pain of a year's duration in the knee joint, and 25 % have severe and disabling knee pain (Jinks et al., 2004). Studies in the United States have shown that 13.9% of adults aged 25 or over have osteoarthritis, and of those people over 65 years old, there are 33.6% with knee OA, with about 27 million persons of all ages in America having the disease (Lawrence et al., 2008).

B. *Gender*:

Knee OA incidence is higher in women. In 2015, a systematic review was done for the population aged over 50 and reported that 11 cohort studies found being female to be a possible risk factor (Silverwood et al., 2015). Knee OA symptoms presented in women over the age of 50 and increased dramatically at about menopause time (Andriacchi, 1994, Felson et al., 2000, Srikanth et al., 2005), while it was higher in men in the age group younger than 50 years old (Felson et al., 2000). In addition, knee OA is more common in males younger than 45 and women aged 55 years or over (Silman and Hochberg, 2001, Iqbal et al., 2011).

C. *Hormone effect:*

Oestrogen regulates bone metabolism, and OA increases dramatically in women in the years after the menopause, due to lower levels of the oestrogen hormone (Felson and Nevitt, 1998). Pre-menopausal women have greater risk of OA development as the hormone raises bone mass, which increases the load on the cartilage (Nevitt et al., 1996).

D. Ethnic differences

Knee OA prevalence varies among different ethnic and racial groups. Thus, the knee osteoarthritis risk in non-Hispanic white women is lower as compared with people of African–American descent (Felson and Nevitt, 1998).

E. *Vitamin deficiency*:

A subclinical deficit in vitamin K was correlated with raised risk of progression of knee OA from results shown in radiographic and MRI assessments of cartilage lesions (Misra et al., 2013). Also, Zhang et al. (2014) suggest that people who have a deficiency in vitamin D have a raised incidence and risk of knee osteoarthritis (Zhang et al., 2014).

2.2.2 Environmental/Biomechanical factors:

A. Obesity:

People who are obese or overweight have a high prevalence of osteoarthritis in the knee joint. Further, in people who are suffering from knee osteoarthritis, being overweight raises the risk of radiographic progression of the disease (Dougados et al., 1992, Schouten et al., 1992). In addition, it is clear from previous studies that obese individuals have a high risk of progression in knee OA (Cooper et al., 2000, Felson and Nevitt, 2004). So, obesity is a significant factor in increasing knee OA progression (Felson et al., 1987) and risk significantly increases, from 9 to 13%, for each kg of body weight increased (Cicuttini et al., 1996). In Saudi Arabia, obesity is extremely widespread and may lead the Kingdom of Saudi Arabia to have one of the highest incidences of knee osteoarthritis in the world (Al-Arfaj and Al-Boukai, 2002).

B. Previous knee injury:

Knee injury is one of the most presently documented risk factors in knee OA (Cooper et al., 2000). The population with joint injuries, including articular surface fractures, have a raised percentage of knee joint OA. Ligaments and menisci tears (Honkonen, 1995): for example, meniscectomy after knee injury; were radiographically shown as a considerable risk factor for knee OA, with an association found twenty-one years after meniscectomy (Roos et al., 1998). Moreover, ACL tears in soccer-playing women of 12 years' duration showed high radiographic incidence of OA in the knee joint (Lohmander et al., 2004). In addition, a recent systematic review including meta-analysis reports that 13 cohort studies show previous injury of the knee joint as a risk factor for the start of osteoarthritis in the knee (Silverwood et al., 2015).

C. Occupational-related joint stresses:

An increased risk factor, from moderate to severe, for radiographic knee osteoarthritis is associated with long knee-bending activities such as squatting, and repetitive knee stresses (Muraki et al., 2009, Cooper et al., 1994). In the Framingham study, it was suggested that from 15% to 30% of knee OA is due to occupational activity such as stair climbing, walking on uneven ground, or repeated knee bending and standing for prolonged periods (Felson et al., 1991, Cooper et al., 1994).

It is clear from the above discussion that a range of intrinsic and environmental factors can contribute both to the onset and progression of OA of the knee. These include age, gender, ethnic variation, obesity, previous knee joint injury and occupational-related stresses. A range of different biomechanical factors also have the potential to contribute to knee joint loading and therefore to the onset and progression of knee OA. These factors can only be measured using complex laboratory testing and so are not typically incorporated into large-scale cohort or case-control studies. Nevertheless, these factors may provide important insights into the aetiology of knee OA. In the following section, a short summary of biomechanical measurement is presented, followed by a detailed discussion of the specific biomechanical factors that have been linked with knee OA.

2.3 Measuring the biomechanics of walking

Definition of biomechanics

The term biomechanics contains two parts; the prefix word *bio*, and the root, *mechanics* (Hoffman, 2009). *Bio* pertains to the living, while *mechanics* pertains to the action of forces on the body which result in balance and motion. So, 'biomechanics' is the scientific knowledge of the living body's movement, including bones, tendons, ligaments and muscles, which work with each other to produce movement. Therefore, biomechanics can be clearly defined as the study of motion of the living organism, such as the human, using mechanical science (Hatze, 1974). Newtonian mechanics as applied to the skeleton, nerves and muscles of the body is known as biomechanics, and is based on the forces applied which lead to motion, and the internal forces that apply within the body (Rau et al., 2000, Rose et al., 2006). Biomechanics provides the researcher with the necessary tools (conceptual and mathematical) to understand how the movement of living things is performed (Knudson, 2007).

2.3.1 Gait cycle:

The gait cycle or walking cycle is defined as the period of time that starts when one foot touches the ground and finishes when the same foot touches the ground again during human walking. This gait cycle must contain two steps to complete one cycle. The gait cycle has two basic phases: the stance phase and the swing phase:

- *Stance phase*, in which the foot is in touch with the ground.
- *Swing phase*, in which the foot is not in touch with the ground (in the air) and the limb swings forward.

The stance phase can be divided into single limb support and double limb support in general within the gait cycle (Perry and Davids, 1992). Moreover, there are eight subdivisions during the full gait cycle, with five of these in the stance phase (about 60% of the gait cycle). These are: 1) initial contact; 2) loading response; 3) mid stance; 4) terminal stance; and 5) pre-swing. The remainder are in the swing phase, forming approximately 40% of the gait cycle: 1) initial swing; 2) mid swing; and 3) terminal swing, as shown in the figure below:

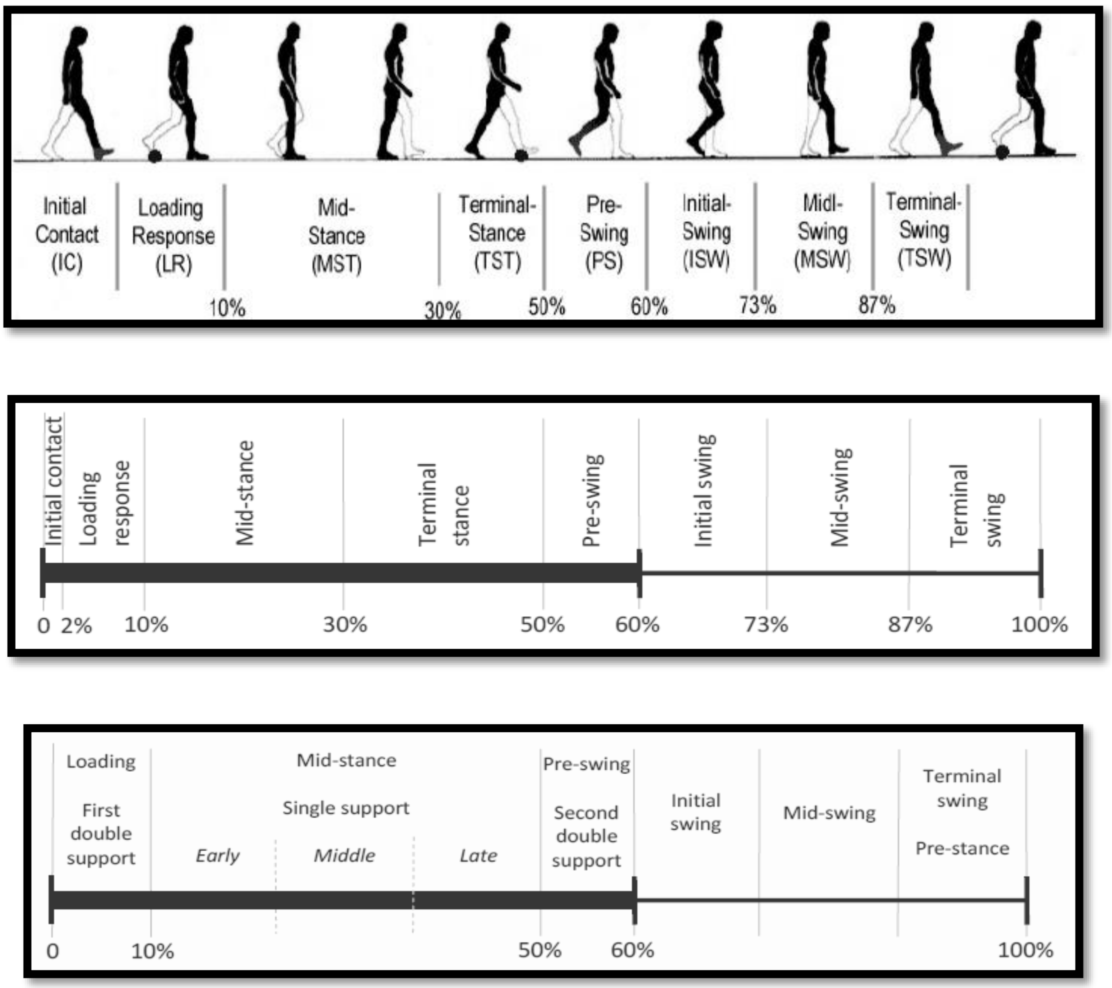


Figure 2-2. The phases of the gait cycle in walking (Adapted from Perry, 1992).

2.3.2 Kinematics & Kinetics

Kinematics and kinetics are widely used in gait analysis. Kinematics describe the speed, extent, and direction of movement of the joint, however, kinematics descriptions do not provide insight into causes of motion. For this, it is necessary to look to kinetics which is the branch of biomechanics concerned with the causes of motion through forces and joint moments (Richards, 2008; Perry et al., 1992). The joint moment is the turning force around the joint centre, created by the action of the different muscles and the effect of gravity. The moments at different phases of the gait cycle can be understood from a consideration of position and direction of the ground reaction force (GRF) vector relative to the position of the joint under study. This idea is illustrated below (Levangie and Norkin, 2011). For example, during the first half of stance, the GRF vector first passes in front, then behind and then again in front of the knee joint centre. This gives rise to the pattern of a flexor, extensor and then again a flexor moment, as illustrated below.

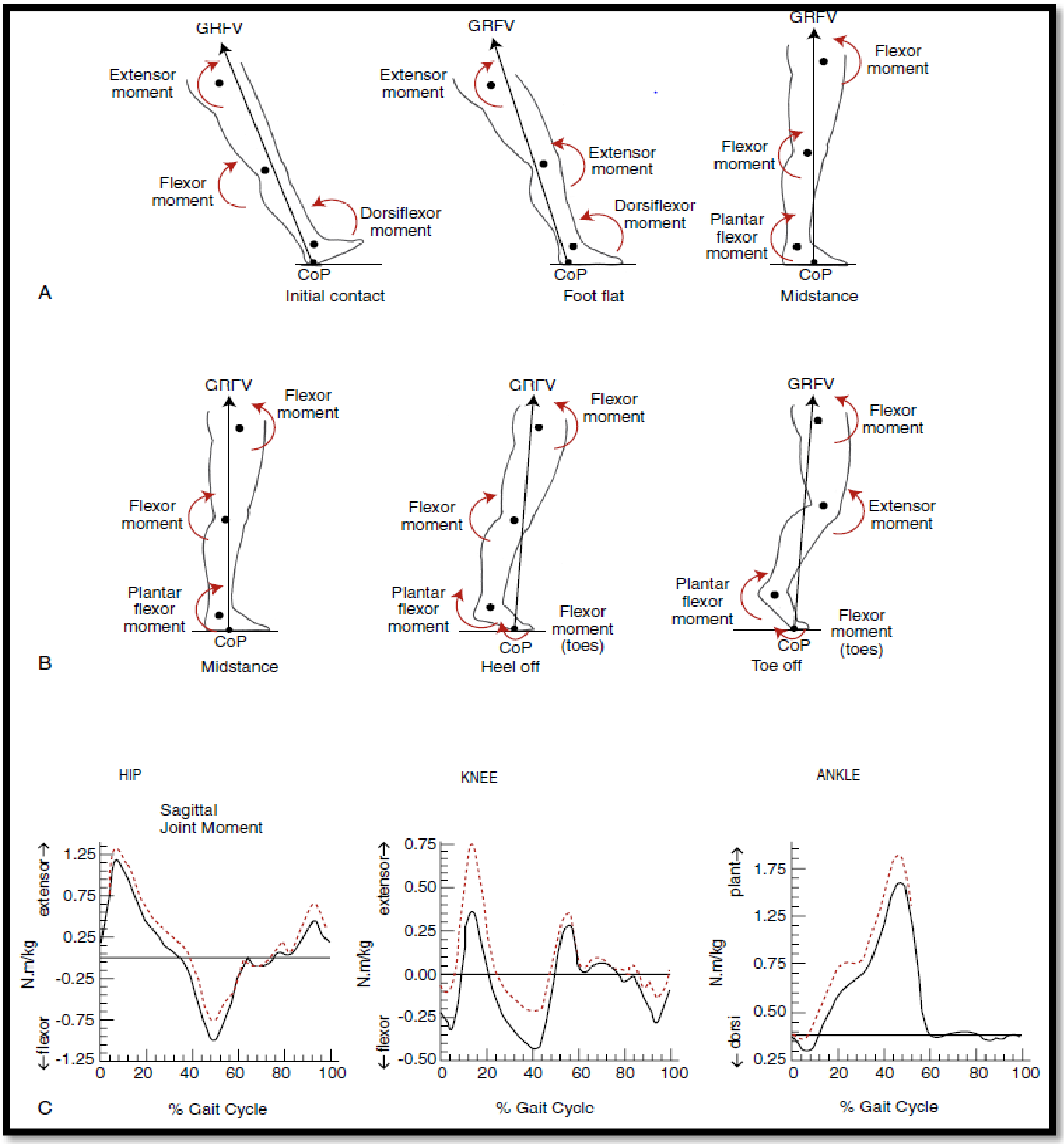


Figure 2-3. Patterns of internal moments in the sagittal plane at the hip, knee, and ankle with centre of pressure (CoP) and ground reaction force vectors (GRFVs). The dotted lines represent the standard deviations, and the solid lines represent the mean values. Reproduced from Levangie and Norkin (2011).

2.3.3 Measuring muscle activation using EMG

Surface electrodes recording kinesiological electromyographic (EMG) data are commonly used to measure muscular activity during walking and determine normal and pathological motor strategies (Frigo and Crenna, 2009). The myo-electric signal acts as a helpful indicator of muscular mechanical effect. The amplitude of the EMG signal obtained during walking may be translated as a measure of relative muscle activity (Kunju et al., 2009). The timing of the EMG during each phase or the whole gait cycle gives information about muscle integration and neurological control (Perry et al., 1992). Although EMG amplitude is straightforward to quantify, this parameter needs to be interpreted with caution, as the magnitude of EMG signals, as well as reflecting electrical impulses, is also influenced by how conductive the tissues are which separate the muscle and the electrodes on the skin. Further, the fact that the force of the muscle and electrical impulses are not connected in a linear manner means that it is important to be cautious in linking the amplitude of the EMG signal to the muscle forces associated with a dynamic movement (De Luca, 1997).

2.4 Previous research into biomechanical differences between knee OA and control groups

2.4.1 Differences in spatiotemporal gait parameters and kinematic characteristics

It has been reported that patients with knee osteoarthritis walk significantly more slowly than healthy subjects (Andriacchi et al., 1977, Chen et al., 2003). Previous studies have also demonstrated alterations in spatiotemporal gait variables in those with different severities of knee osteoarthritis (Al-Zahrani and Bakheit, 2002, Kaufman et al., 2001). People who suffer from osteoarthritis of the knee, mainly in the medial compartment of the joint, have slower walking speeds (Kaufman et al., 2001), shorter step lengths, longer time of double support, reduced length of stride and reduction in cadence (Al-Zahrani and Bakheit, 2002). In addition, people with knee OA have been shown to have a prolonged stance phase (Al-Zahrani and

Bakheit, 2002, Landry et al., 2007, Astephen et al., 2008) when compared with healthy control subjects. A slower walking speed in knee OA sufferers has been suggested to decrease loading on the knee joint (Mündermann et al., 2004). However, it is also possible that the alterations in spatiotemporal gait parameters may be a strategy to increase the body's stability of centre of mass (CoM) during walking.

Studies comparing knee joint kinematic parameters between healthy individuals and those with knee OA often demonstrate contradictory findings. For example, some studies demonstrate greater knee flexion angles at initial contact (Baliunas et al., 2002a, Childs et al., 2004), while others show greater knee extension angles (Rudolph et al., 2007, Smith et al., 2004). Similarly, during early stance, peak flexion angle at the knee joint also varies between studies, with greater knee flexion for controls (Al-Zahrani and Bakheit, 2002, Lewek et al., 2006) or no variation in peak knee flexion values (Baliunas et al., 2002b). Several studies of knee OA patients have reported reductions in knee flexion excursion (Childs et al., 2004, Lewek et al., 2006, Rudolph et al., 2007), while other studies have shown patients with OA to exhibit greater knee flexion angles at initial contact, and during early and late stance (Heiden et al., 2009b). These contradictory findings most likely indicate varied kinematic responses to the disease. However, as explained above, kinematic patterns give no indication of the underlying forces and so provide minimal insight into differences in joint loading between people with knee OA and healthy controls.

2.4.2 Differences in joint moments

Joint moments can be considered as either external or internal. External moments are created about the centre of the joint by external forces such as gravity. External moments oppose internal moments and are developed via muscle and soft tissue forces (Baliunas et al., 2002b).

Internal net joint moments are those required by the body to resist the external forces applied. For example, when the ground reaction force vector passes posterior to the knee, creating a knee flexion moment, the quadriceps or knee extensor muscles must contract to create a knee extension moment to resist knee flexion. Net internal joint moments reflect muscle activity and can be used to obtain insight into which particular muscles are active during a specific phase of movement. Furthermore, joint moments can be used as an indicator of joint loading and therefore are often used in the study of injury and injury prevention (de David et al., 2015). The paragraphs below highlight the key differences in joint moments which have been observed between healthy participants and individuals with knee OA.

Frontal plane

The term knee adduction moment is used to describe the frontal plane moment at the knee joint and provides a useful measure of the load distribution between the medial and lateral compartments of the knee joint (Hurwitz et al., 2002, Henriksen et al., 2006). The knee adduction moment is determined as the product of the magnitude of the ground reaction force (GRF) and the frontal moment arm at the knee (perpendicular distance from the knee joint centre to the GRF) (Maneekittichot et al., 2013) and has been widely studied in knee OA. For example, this biomechanical parameter was examined in patients with knee OA and matched healthy controls by Mundermann et al. (2005). The knee adduction moment was found to be elevated at the first peak (during mid-stance) and the second peak (during terminal stance) in patients with severe knee OA, while in patients with early knee OA, the second peak was lower (Maneekittichot et al., 2013). Thorp et al. (2007) reported that, in patients with Kellgren–Lawrence grade 2, the adductor moment of the knee joint increased by 10 % compared to healthy persons (Baliunas et al., 2002b).

Sagittal plane

In addition to alterations in frontal plane moment, previous research has observed kinetic alterations in the sagittal plane associated with knee OA. Several studies comparing individuals with knee osteoarthritis and healthy controls have shown that, during mid-stance, knee flexion moments are decreased (Kaufman et al., 2001, Astephen et al., 2008) (see Figure 2-4 below). This gait pattern in knee OA patients may have been adopted in an attempt to decrease joint loading in order to relieve pain (Baliunas et al., 2002a) and may be achieved through altered upper body positioning and/or altered muscle activation patterns. Interestingly, other studies report minimal differences in knee moments during this phase of the gait cycle (Sritharan et al., 2016; Mündermann et al., 2005). However, it is not clear whether these contradictory findings are the result of differences in severity of OA between the different studies.

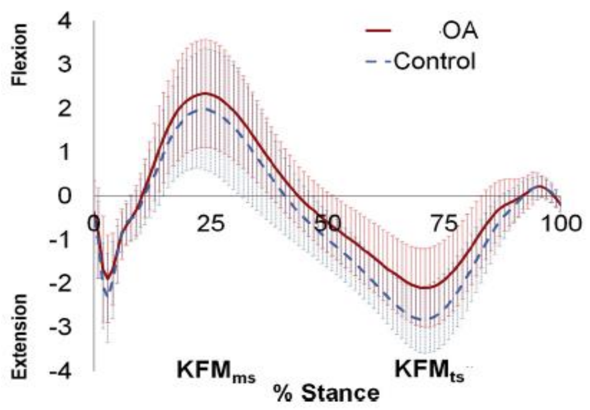


Figure 2-4. Sagittal plane knee moment for healthy subjects (blue dashed), OA patients at baseline (red solid) adapted from (Edd et al., 2017)

Other studies have investigated potential differences in hip and ankle moments in the sagittal plane during walking between OA and control (Liu et al., 2014, Judge et al., 1996, Apps et al., 2016, Chien et al., 2014). Although the patterns of ankle moments are typically similar between healthy individuals and those with knee OA, there are subtle differences in the pattern of hip extensor moment during mid-stance (Liu et al., 2014). These differences were highlighted in a previous study comparing knee OA with healthy controls (see Figure 2-5 below) (Liu et al., 2014). Although peak hip moments were similar between these two groups, the subjects with knee OA appeared to walk with greater hip extensor moments in mid-stance and the time at which the hip moment changed from an extensor moment to a flexor moment was delayed.

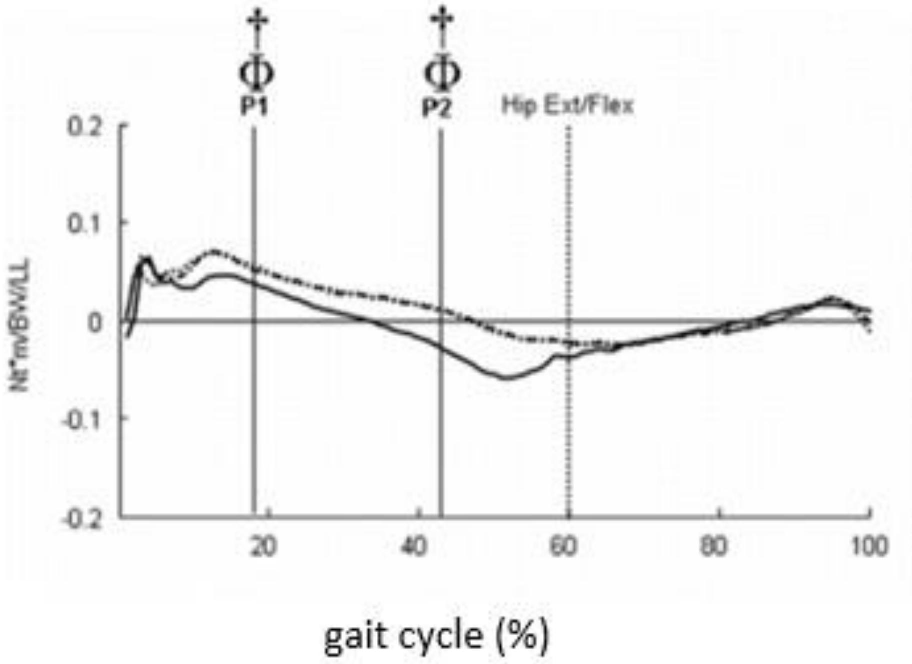


Figure 2-5. Mean curves of the hip joint moments in the sagittal plane, for the control (solid line) and the OA group (dotted line) (Liu et al., 2014).

In summary, previous research has demonstrated clear differences in frontal plane moments in people with knee OA when compared to healthy individuals. However, there are also differences in sagittal plane moments. These differences are characterised by a reduced extensor moment at the knee and an increased hip extensor moment during early and mid-stance.

2.4.3 Differences in muscle activation and muscular co-contraction

To date, there has been a large number of studies which have investigated differences in muscle activation between healthy individuals and those with knee OA. These studies tend to report co-contraction, which is defined as the simultaneous activity of synergistic muscles (agonist and antagonist) (Sirin and Patla, 1987). Over recent years, research has shown that muscle co-contraction is increased in individuals with OA compared to healthy subjects (Childs et al., 2004, Lewek et al., 2004a, Hubley-Kozey et al., 2006). It has been suggested that muscle co-contraction in patients with knee OA will increase joint loading at the medial compartment (Andriacchi, 1994). This idea has motivated a substantial number of studies comparing muscle co-contraction between patients with knee OA and healthy subjects (Childs et al., 2004; Lewek et al., 2004b; Hubley-Kozey et al., 2006). This research is outlined below.

In general, research has shown increased activity of the hamstrings and quadriceps (Zeni et al., 2010, Childs et al., 2004, Hortobagyi et al., 2005, Hodges et al., 2016). For example, Zeni et al. (2010) investigated the muscle co- contraction of hamstring and quadriceps in people with medial knee osteoarthritis. In this study, they recruited two groups; healthy and medial knee OA groups. The knee OA patients were divided into three groups based on severity of disease to investigate whether previously observed differences in co-contraction changes were due to walking speed or associated with severity of OA. EMG data were analysed from the vastus lateralis and semimembranosus of one leg and normalized by MVIC (maximal voluntary isometric contraction) using data collected as subjects walked at three different speeds; self-selected over the ground, 1.0 m/s, and the fastest speed possible on the treadmill (Zeni et al., 2010). This study showed higher co-contraction in the knee OA groups with severe and moderate disease in comparison to the healthy subjects when walking at 1.0 m/s, illustrating that knee OA is associated with higher co-contraction regardless of speed or the severity of OA.

In addition to quadriceps and hamstring activity, a small number studies have also investigated gastrocnemius activity (Schmitt and Rudolph, 2007, Childs et al., 2004, Sritharan et al., 2016a). This research shows that this muscle is often overactive during walking in individuals with OA (Schmitt and Rudolph, 2007, Childs et al., 2004, Sritharan et al., 2016a). In addition to functioning to plantar flex the ankle, the gastrocnemius muscle also acts to flex the knee joint. Therefore, an increased plantar flexor moment may reduce the net knee extensor moment and may explain the results of previous research that has shown that people with knee OA tend to walk with lower peak knee extensor moments in early stance (Huang et al., 2008, Kaufman et al., 2001).

In general, studies investigating differences in muscle activation between healthy individuals and those with knee OA have shown increased quadriceps, hamstrings and gastrocnemius activity (Hortobagyi et al., 2005, Childs et al., 2004). These increases in muscle activation give rise to increased co-contraction, which, as explained below, will increase the compressive loading on the knee joint.

2.4.3 Current models to explain co-contraction.

The benefits and drawbacks of co-contraction are still subject to controversy. Lewek et al. (2005) evaluated co-contraction of antagonist muscles of the upper and lower leg in patients with knee OA, and, following on from Chipplein and Andriacchi (1991), showed that patients with knee OA had higher levels of co-contraction than healthy subjects. Also, there was a positive correlation between better knee stability and higher co-contraction in patients with OA. The authors concluded that patients with medial knee OA try to stabilize the knee with greater co-contraction of the muscles on the medial side in response to slackness that shows on the medial side of the joint. However, Lewek at al. (2005) suggested that may plays a role in higher joint compression and therefore, as it could aggravate joint damage, it should be reduced to slow the progression of knee OA. Nevertheless, in a letter to the journal editor, Steultjens et al. (2006) suggested that the experimental results of Lewek et al. (2005) can be interpreted to mean that antagonist muscle co-contraction is a strategy employed to steady the knee joint in the absence of sufficient stabilization by the passive structure (capsule and ligaments) of the knee. In addition, Steultjens et al. (2006) have interpreted the experimental findings of Childs et al. (2004) to mean that, in the absence of sufficient stabilization by the passive structure, muscle activity becomes even more essential in preserving walking ability in knee OA. However, these ideas are in contrast to those originally proposed by Lewek et al. (2005), who proposed that co-contraction needs to be reduced as it will increase compressive loads across the joint. So, continued research is needed to identify whether increased co-contraction is harmful or helpful in patients with knee OA.

A recently published article by Sritharan et al. (2016) aimed to investigate the influence of muscles on medial knee forces in knee osteoarthritis to compare muscle forces with healthy subjects during gait. The results show that small increases in knee flexor muscle (hamstring and gastrocnemius) and knee extensor muscle (quadriceps) activity can lead to significantly higher joint compression at the knee. This study highlighted the complex relations between gravity, muscle activation and knee joint loading (Sritharan et al., 2016a). Interestingly, the study showed that subtle alterations in the activation of specific knee muscles during gait, such as a decreased muscle activity in the hamstrings, quadriceps, and gastrocnemius, significantly increase compressive force at the knee joint during gait (stance phase). Such increases in compressive loading have the potential to both increase pain and also to increase the rate of cartilage loss and therefore accelerate disease progression. These ideas are discussed in more detail below.

2.4.4 Co-contraction and disease progression

To date, there have been two studies investigating the link between co-contraction and the rate of knee OA disease progression. In one study, Hodges et al. (2016) collected data from 50 patients with medial knee osteoarthritis. The cartilage of the medial compartment of the knee was measured at baseline and again at one year using MRI techniques and changes in cartilage thickness were related to muscle activation patterns, measured using EMG. The authors found that the people who had higher co-contraction at baseline demonstrated a more rapid loss of cartilage, indicating that muscle co-contraction accelerates the progression of knee OA (Brandon et al., 2014). Based on this finding, the authors suggested that research is needed into biomechanical interventions which may be effective at changing patterns of knee muscle activation. This may offer a possible way to decrease joint loading, and consequently, the rate of progression of knee OA (Brandon et al., 2014).

In another study, Hubley-Kozey et al. (2013) examined fifty patients with medial knee OA of a moderate level of severity. Patients were then followed from baseline for eight years, with 50% choosing to undergo total knee replacement (TKR). Baseline measures of co-contraction and muscle strength were obtained and used to understand risk factors for TKR. The study found that the patients who chose to have total knee arthroplasty (TKA) had higher co-contraction in the knee flexor muscle (hamstrings) and knee extensor muscle (quadriceps) during the mid-stance of the gait cycle. This muscle activity pattern in total knee replacement patients is consistent with the idea of co-contraction increasing knee joint loading and accelerating knee OA progression through increased compressive joint loads (Hubley-Kozey et al., 2013). Together, these two studies show that increased co-contraction has the potential to accelerate disease progression. This is most likely because increased co-contraction will increase the compressive loads at the knee joint. Modelling studies have the potential to provide insight into how co-contraction can elevate joint loads and are therefore discussed below.

2.4.5 Mechanical knee joint loading in knee OA

A recent study provides useful insight into mechanical loading at the knee during walking (Brandon et al., 2014). This study used a modelling approach to understand how muscle activation patterns, typically observed in people with knee OA, may affect joint loading on the lateral and medial compartment of the knee joint during gait. The plots shown in Figure 2-6 below illustrate the medial compartment load (A), lateral compartment load (B) and total load (C), and show that there are two clear peaks in the medial load which occur at approximately 10-15% of the gait cycle and at around midstance. Interestingly, the model showed that changes in muscle activation characteristic of knee OA (shown in red below) led to increases in contact force: especially around the initial peak in loading. Furthermore, this paper showed that selective activation of lateral knee muscles (vastus lateralis), as found in subjects with medial knee OA, would not independently reduce medial knee contact loads as has been suggested by some researchers (Hodges et al., 2016).

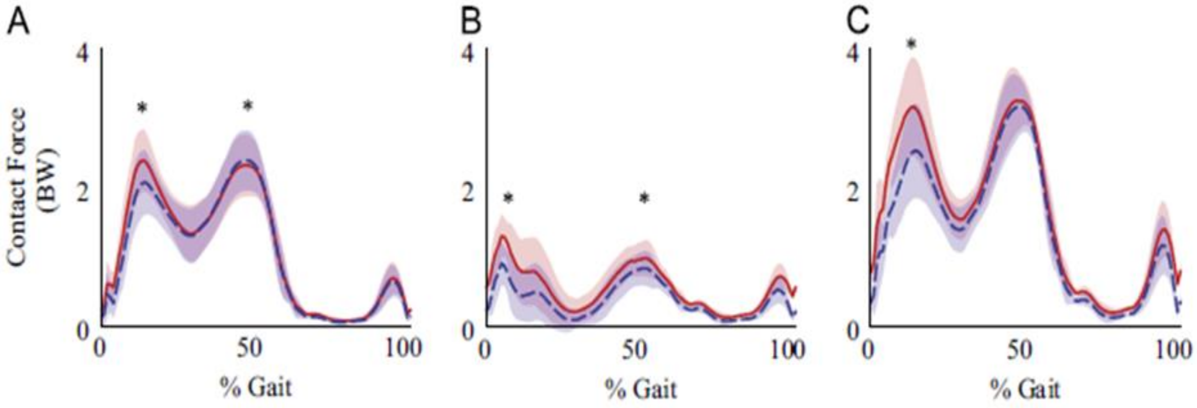


Figure 2-6. Mean & standard deviation (shaded) (A) medial, (B) lateral and ((C) total, sum) axial knee contact force during normal gait in eight subjects with moderate knee osteoarthritis. Forces predicted for "Baseline" condition (blue, dashed) were lower than those predicted after applying an "OA-type" activation perturbation (red, solid) to vastus lateralis, biceps femoris (LH), and medial Gastrocnemius (Brandon et al., 2014).

In a similar study, Sritharan et al. (2016) used a modelling approach to understand the contribution of three factors to medial contact forces; the muscles which span the knee, the muscles which do not span the knee, and gravity. Their results again illustrate that OA type activation patterns (shown in red) will result in elevated joint contact force at the medial compartment (Sritharan et al., 2016a). However, the exact pattern of increase is slightly different to that reported by Brandon et al. (2014). These differences may be due to the precise nature of the models. Whereas Brandon et al. (2014) sought to understand the effect of changing muscle activation patterns on joint loading. Sritharan et al. (2016) calculated joint loads from a cohort of healthy and a cohort of knee OA participants. However, despite these subtle differences in results, both studies clearly illustrate that elevated hamstring, quadriceps and gastrocnemius activity will increase medial joint loading and that this loading will be significantly higher at approximately 10-15% of the gait cycle (15-25% stance), close to the initial peak in loading.

Selection of outcome

The two studies discussed above provide important insight into the effects of increasing co-contraction on the compressive forces on the knee joint. In the first study, Brandon et al. (2014) showed that if muscle patterns change from those characteristic of a healthy person (low co-contraction) to those more characteristics of somebody with knee OA (high co-contraction), then there is a corresponding increase in the knee contact force, in both the medial and lateral compartment. The figure below shows that peak medial knee joint loading occurs at two distinct phases of the gait cycle. However, changes in muscle activation produce the most pronounced increase in joint force during the first period, with minimal change at the second peak. This first peak occurs at 12.5% of the full gait cycle, at time point which corresponds to 20% of stance, assuming stance to last for 62% of the gait cycle. Given this observation, all subsequent analysis in this thesis are focused on the time period 15-25% of stance, a windows which is centred on the timing of the peak in knee loading most affected by increased co-contraction.

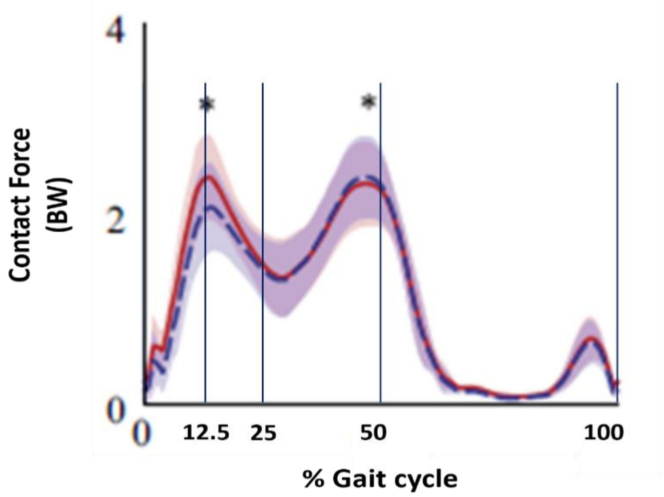


Figure 2-7: Contact knee force during normal gait with low co-contraction (blue) and high co-contraction (red).

2.5 A new model to explain altered moments and increased co-contraction in people with knee OA

In the sections above, the findings of previous research investigating joint moment, muscle activation patterns and joint loading in people with knee OA were reviewed. These studies have identified three important characteristics of knee OA walking gait, which are:

1. Moments: peak knee extensor moments are reduced but hip extensor moments increased during midstance in people with knee OA.
2. Muscle activity: hamstring activity, quadriceps activity and gastrocnemius activity are increased in people with knee OA.
3. Joint loading: elevated levels of muscle activity, characteristic of knee OA, lead to increased compressive loading between 15-25% of stance phases during walking.

Although previous authors have argued that increased co-contraction is an appropriate coping strategy adopted by people with knee OA to stabilise the knee joint, others suggest that it may be maladaptive response to the disease as it leads to increased knee joint compressive forces. In this section we propose an alternative model which may explain both the alterations in joint moment and the increases in muscle activity, characteristic of people with knee OA. This model is based around a biomechanical understanding of increasing the forward lean or inclination of the trunk and is described in detail below.

2.5.1 Trunk inclination and altered joint moments

The plot below illustrates the difference in hip moments between healthy subjects (solid line) and those with knee OA (dashed line) (Liu et al., 2014). Although there is minimal difference in the peak hip extensor moment, which occurs just after initial contact, during the remainder of stance (10-60% of gait cycle), the hip extensor moment is increased in people with knee OA and there is a delay in the time at which the hip moment transitions from an extensor to a flexor moment.

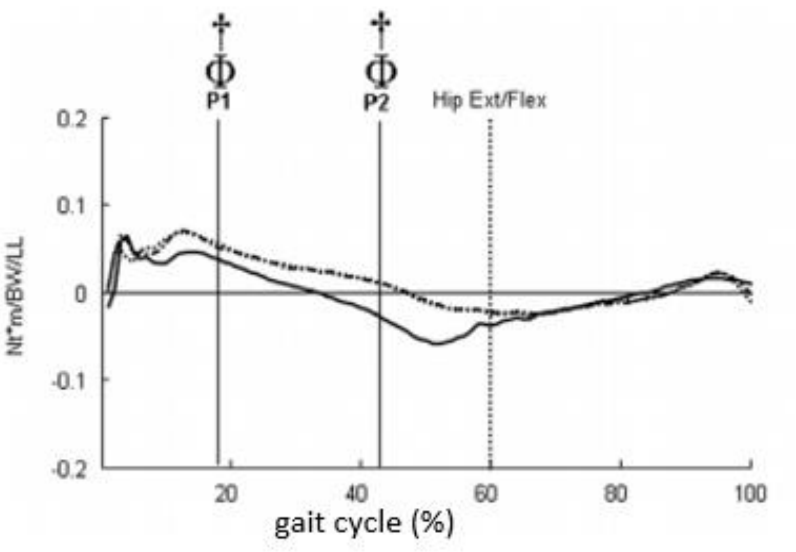


Figure 2-8. Mean curves of the hip joint moment in the sagittal plane, for control subjects (solid line) and OA individuals (dotted line) (Liu et al., 2014).

Interestingly, the pattern shown above is similar to that observed in a study which compared hip flexor moments between individuals who walk with different amounts of trunk forward lean (Leteneur et al., 2009). This study recruited a cohort of 25 individuals who were then divided into two groups based on their natural trunk inclination during walking. The first group (the backward leaners) had a trunk lean of -1.7°, while the second group (the forward leaners) had a trunk lean of 2.9°. The data showed a clear difference in hip moments between these two groups (see Figure 2-9 below), despite a mean difference in trunk inclination of less than 5° between the groups.

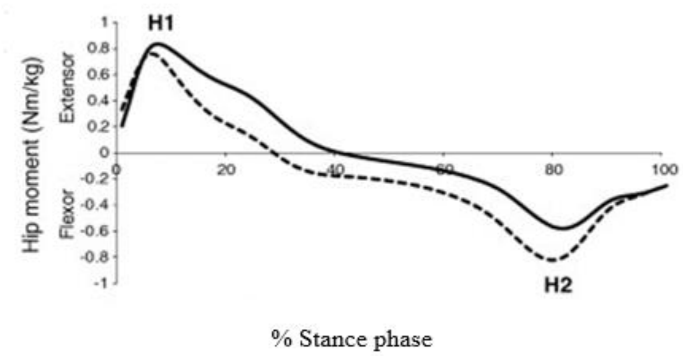


Figure 2-9. Hip moments for the right limb as function of the stance phase for the backward trunk inclinations (dotted line) and forward trunk inclinations (solid line) (Leteneur et al., 2009).

The plot above shows that, in subjects who habitually walk with a forward lean, hip extensor moments are elevated from early stance through to midstance, although interestingly, the peak hip extensor moment is unchanged. These increased moments are likely to be a response to the demand of having to support the trunk segment against gravity, which will require an increased extensor moment at the hip. The similarity between this data and the figure above, displaying differences, between healthy and knee OA, suggests that differences in joint moments previously observed in people with knee OA may be explained by a forward inclination of the trunk. Interestingly, the study investigating differences in moments associated with trunk lean (Leteneur et al., 2009) also showed a small decrease in knee moment and a small increase in ankle moment to be associated with forward lean. Again, these patterns are similar to those reported in people with knee OA (Zeni and Higginson, 2011).

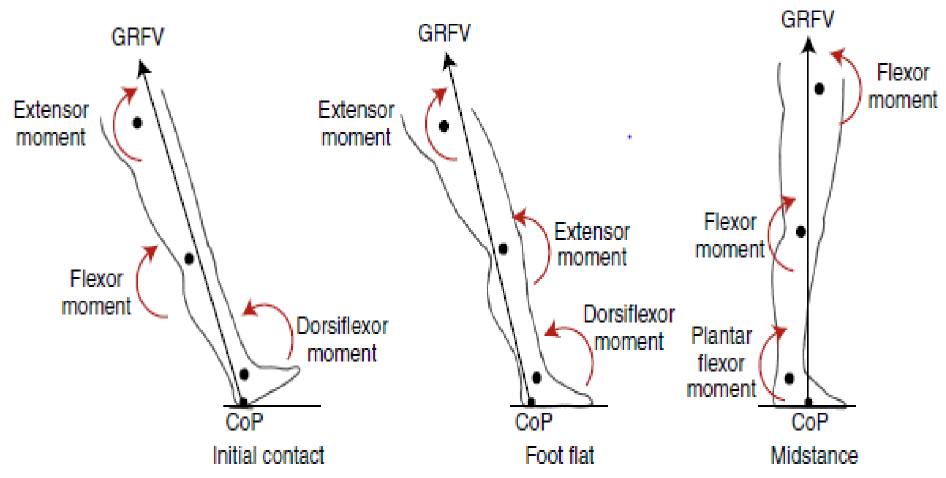


Figure 2-10. Patterns of internal moments in the sagittal plane at the hip, knee, and ankle with centre of pressure (CoP) and ground reaction force (GRF) vectors. The dotted lines represent the standard deviations, and the solid lines represent the mean values. Reproduced from Levangie and Norkin (2011).

To better understand why these changes in moment occur, it is helpful to consider the changes in the direction of the GRF vector which would accompany a change in trunk inclination. The figure above illustrates the orientation of the GRF vector during the early stance and midstance phases of the gait cycle. The head-arms-trunk (HAT) segment accounts for approximately 65% of body mass (Dempster, 1955) and therefore, small changes in the inclination of the segment have the potential to influence the position of the centre of mass (CoM) and therefore the direction of the GRF vector. With an increase in forward lean, there will be a relative anterior movement of the GRF vector relative to the hip joint and this will lead to an increase in the hip extensor moment. Similarly, if there is no change in knee flexion-extension angles, then the anterior movement of the GRF relative to the knee joint (which would result from increased forward lean) would decrease the knee extensor moment (or increase the knee flexor moment).

In order to understand the effect of forward lean on ankle moments, it is necessary to consider the possible effect of increased forward lean on the CoP position. There are two possible scenarios. Firstly, increased forward lean leads to an anterior shift in the position of the CoP as body weight is moved onto the forefoot region. In this scenario, there would be an increase in the ankle plantarflexor moment due the anterior shift on the CoP relative to the ankle joint. In the second scenario, forward lean is not associated with a change in the CoP and therefore there is no corresponding change in the ankle moment.

The changes in joint moments which are characteristic of people with knee OA (detailed in Section 2.4.2) appear, to some degree, to be consistent with a gait pattern in which an increased forward lean has been adopted. Specifically, previous studies have reported increased hip moments (Huang et al., 2008) and decreased knee moments (Debbi et al., 2014). However, most studies report relatively little change in the ankle joint moment (Astephen et al., 2008) suggesting that the CoP is not always shifted anteriorly. However, it is possible that the slower walking speed often adopted by patients with knee OA (Al-Zahrani and Bakheit, 2002, Zeni and Higginson, 2011, Andriacchi et al., 1977, Kaufman et al., 2001) would lead to a corresponding reduction in the vertical GRF. This may offset changes in the position of the CoP and show similar ankle moments. Interestingly, a recent study found that a cohort of patients with knee OA did exhibit a more anterior-shifted CoP during walking (Saito et al., 2013). However, in this study, controls walked faster than the people with knee OA and this could explain the observed differences in CoP pattern.

Changes in joint moments will be accompanied by corresponding changes in muscle activation patterns, as external moments which result from the GRF will need to be balanced by internal moments created by the muscles. It is therefore interesting to map out possible changes in muscle activity which could accompany the changes in joint moments describe above. This is described below.

2.5.2 Trunk inclination and hamstrings-quadriceps co-contraction

A large number of studies have demonstrated increased co-contraction between the hamstrings and quadriceps (Zeni et al., 2010; Childs et al., 2004; Hortobagyi et al., 2005) in people with knee OA (Heiden et al., 2009b). In the section above, a mechanism was proposed for how increased trunk inclination could lead to increased hip extensor moments. During walking, the hamstrings function to both extend the hip joint and also to flex the knee joint. Generating an increased hip extension moment will require increased hamstring activity. However, given the two-joint nature of this muscle group, increased hamstring activity will also increase the flexor moment at the knee. Therefore, it is possible that, in order to balance this increased hamstring activity at the knee, there will also be an increase in quadriceps activity. This increase in both hamstring and quadriceps activity will result in elevated co-contraction. Peak compressive knee joint loading occurs at approximately 15-25% of stance phase (see Section 2.5). Interestingly, during this period, hip extensor moments are elevated considerably in individuals who walk with an inclined trunk position (Leteneur et al., 2009) and consequently hamstring activity is also likely to be increased. Therefore, the possibility exists that co-contraction between the hamstrings and quadriceps, and therefore joint loading, results from a forward inclination of the trunk.

2.5.3 Trunk inclination and gastrocnemius-quadriceps co-contraction

Research has shown that individuals with knee OA walk with elevated levels of gastrocnemius-quadriceps co-contraction (Childs et al., 2004, Vieira et al., 2010). These studies have quantified gastrocnemius activity at different parts of the gait cycle, with one study demonstrating greater gastrocnemius activity during the weight acceptance phase (Schmitt and Rudolph, 2007) and another showing an earlier onset of gastrocnemius activity (Childs et al., 2004). Interestingly, a recent study demonstrated increased gastrocnemius activity between 15-25% of stance phase (Sritharan et al., 2016b). As explained in the sections above, this is the period which corresponds to the first peak in compressive loading at the knee joint.

In scenario 1 above, a mechanism was proposed in which an increase in trunk inclination resulted in an increase in both sagittal hip moment and also sagittal ankle moment. This increase in ankle moment resulted from an anterior shift in the CoP position. There are two primary muscles which act to plantarflex the ankle joint: the gastrocnemius and the soleus muscle. Therefore, if an individual responds to a forward trunk inclination with an anterior shift in CoP (as outlined above), then there is likely to be a corresponding increase in gastrocnemius activity. Similar to the hamstring muscles, the gastrocnemius is a two-joint muscle which acts both to plantarflex the ankle and flex the knee. Therefore, if increased gastrocnemius activity does result from an increased in the ankle plantarflexor moment, this may lead to increased quadriceps activity, which would be required to prevent an undesired change in the flexion-extension moment at the knee. Thus, the potential exists for increased gastrocnemius-quadriceps co-contraction to result from an increased inclination of the trunk, provided this inclination is associated with a change in the CoP and therefore the ankle moment.

Modelling studies have identified that a peak in joint loading occurs at around 15-25% of the stance phase of walking. During this period of the gait cycle, sagittal hip moments appear to be elevated in individuals with knee OA (Liu et al., 2014) and sagittal knee moments appear to be reduced (Astephen et al., 2008). Furthermore, EMG activity also appears to be elevated in the hamstrings, gastrocnemius and quadriceps, and this results in elevated co-contraction

(Childs et al., 2004, Hortobagyi et al., 2005, Zeni et al., 2010, Hubley-Kozey et al., 2009, Brandon et al., 2014), and therefore joint loading, during this phase. In the sections above, a model was presented to explain how these alterations in joint moments and muscle activity may have resulted from an increase in trunk inclination. In the following sections, the literature on CoM position and trunk inclination in people with knee OA is reviewed to understand if current evidence supports the idea that people with knee OA walk with a forward trunk inclination.

2.6 Alterations in trunk inclination and centre of mass position in standing and walking in individuals with knee OA

2.6.1 Alterations in trunk inclination in people with knee OA

A recent study by Turcot et al. (2015) studied postural behaviour in people with end-stage (immediately prior to a knee replacement) knee osteoarthritis. By comparing joint positions in standing between their OA group and a healthy group, they were able to demonstrate that when standing, people with knee OA adopt a more flexed posture at all joint levels (see Figure 2-11 below. Interestingly, these data showed OA patients as standing with approximately 3° more trunk inclination that the healthy participants. However, Turcot et al. (Turcot et al., 2015a), did not investigate walking and therefore it is not clear whether the forward inclination of the trunk, adopted in standing is maintained during walking. As far as can be identified, there are no other papers which have reported trunk inclination during walking in people with knee OA.

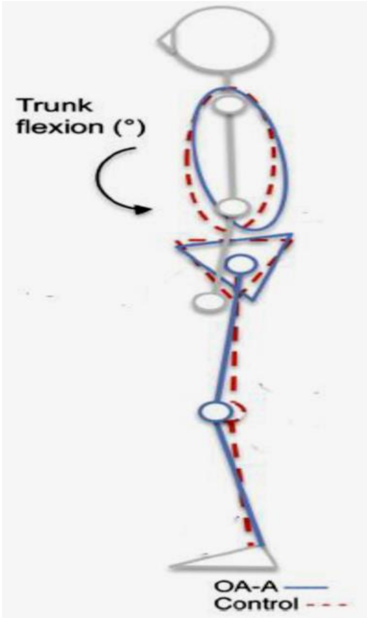


Figure 2-11. Flexed posture in knee OA patients. Illustration of the postural strategy used by the group of patients with knee OA (blue line) and the control group (red dotted line) in the sagittal plane (adapted from Turcot et al., 2015).

2.6.2 Alterations in CoM position in people with knee osteoarthritis

As an alternative to trunk inclination, it is possible that previous research has investigated CoM position and sought to understand whether CoM movements/positions may be altered in people with knee OA. The motion of the CoM reflects whole body motion and therefore can provide insight into whole-body postural control during dynamic movements, such as walking. This idea has motivated previous research which has used the CoM trajectory to understand disturbances for a wide range of gait pathologies (Lord and Menz, 2000, Chen and Chou, 2010, Granata and Lockhart, 2008). This is because alterations in CoM trajectory may point to a clinical manifestation of an underlying pathology or indicate a problem with gait stability. Given this motivation, an exhaustive literature search was performed to identify any studies looking at CoM motions in walking. However, before this work is presented, a brief explanation is given of how CoM position is calculated and also its typical pattern of motion during walking.

Total body centre of mass in the global reference system (GRS) is defined as the weighted average of the CoM of each body segment in 3D space (Winter, 1995) and can be thought of as the imaginary point in a body or system in which its mass is located (Richards, 2008). The location of the human body's centre of mass lies approximately anterior to the second sacrum vertebra (S2) (Figure 2-12). However, CoM location depends on the body weight distribution and relative orientation of each body segment and therefore changes continually during a dynamic movement such as walking. The gold standard method for calculating CoM trajectory during walking is to segment the body into 13 segments (feet, shanks, thighs, pelvis, trunk, head and upper and lower arms) (Halvorsen et al., 2009). With this approach, CoM motion is derived from a weighted average of the positions of the CoM of individual body segments for each frame of kinematic data obtained during the data trial. Anthropomorphic data (e.g. (Dempster, 1955) is used to determine mass fractions and therefore the weighting of the individual segments, and therefore the mean position of the CoM for the whole body.

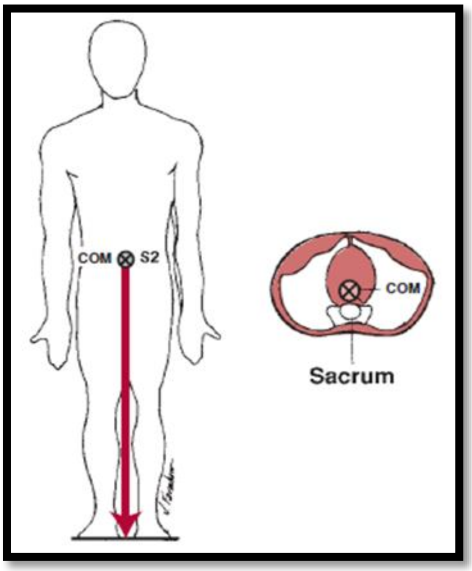


Figure 2-12. The CoM of the human body is located at approximately S2, anterior to the second sacrum (inset). The extended line of gravity lies within the BoS. Adapted from (Levangie and Norkin, 2001).

Two studies have been carried out which have explored differences in CoM trajectory between healthy individuals and people with knee OA (Wang et al., 2010, Hsu et al., 2010). In the first study, Wang et al. (2010) sought to investigate dynamic stability in people with knee OA by studying the position of the CoM with respect to the CoP. A total of 20 subjects were recruited in this study; ten healthy and ten with knee OA, and a full body (12-segment) model was used to calculate CoM position at different parts of the gait cycle. Importantly, they calculated the CoM-CoP inclination angle in the sagittal and frontal plane and compared this between the OA and the healthy group. Their data showed a greater CoM-CoP inclination angle in the sagittal plane during the early- and mid-stance phase, illustrating that the CoM was displaced anteriorly with respect to the CoP in participants with knee OA. However, from the analysis, it is not clear whether this anterior shift of the CoM was the result of an increased trunk inclination.

In the second study Hsu et al. (Hsu et al., 2010) investigated the control of CoM during obstacle crossing in 11 participants with knee OA and 11 healthy control subjects. The researchers used a full body model (12 segments) and also studied CoM and CoP, by quantifying CoM-CoP inclination angles and while obstacle crossing. They reported that the knee OA patients adopted a successful strategy to achieve “better” control of the CoM and with a decreased inclination angle from the sagittal plane (Hsu et al., 2010). However, this study provides limited insight into any possible differences in trunk inclination between healthy and knee OA individuals during normal walking.

The results of the first study, described above, demonstrate that individuals with knee OA walk with an anterior displacement of CoM relative to the CoP. However, it is not clear whether this relative displacement is the result of an increased trunk inclination and/or an alteration in foot placement position. Furthermore, the data presented by Wang et al. (Wang et al., 2010) gives only limited insight into the relative positioning of the CoM with respect to the foot. As outlined above, it is possible that an increased inclination of the trunk may lead to an anterior shift in the CoP. If both CoM and CoP move anteriorly by similar amounts, then their relative displacement could remain unchanged. Thus, reporting only the CoM-CoP distance or angle gives limited insight into the presence or absence of increased trunk inclination. Therefore, although evidence exists to suggest that people with knee OA stand with increased trunk inclination, it is currently not clear whether people with knee OA walk with an inclined trunk.

2.7 Summary and research questions

The gait of people with knee OA is characterised by increased hip extension moments, decreased knee extension moments and increased muscular co-contraction during 15-25% of stance phase. Interestingly, other research has demonstrated that, in people who walk with an increased trunk inclination, there is also increased hip extension moment and a trend towards a reduced knee extension moment. It is therefore possible that the gait characteristics of people with knee OA are a result of an increased inclination of the trunk. However, although previous research has shown that people with knee OA stand with increased trunk inclination, it is not clear whether the increased trunk inclination in standing is maintained during walking. Furthermore, it is not clear whether there is a direct relationship between trunk inclination and joint moments/muscular co-contraction during 15-25% of stance phase in people with knee OA. These limitations in knowledge motivate the first two studies presented in this thesis, which address the specific research questions defined below:

Study 1: Trunk inclination in people with knee OA

This first study focused on characterising the differences between people with knee OA and healthy controls. In addition to exploring differences in trunk inclination during standing and walking (RQ 1A-1C), I also sought to understand differences in CoP and whether these differences could be linked to trunk inclination (RQ1D-1E). In the final part of this study, I sought to characterise differences in moments, muscle activations and co-contraction (RQ 1F-1H), between healthy people and those with knee OA. This comparison was performed to fully characterise the knee OA cohort and to facilitate comparison with previous research. A full list of research questions is provided below.

RQ 1A: Do individuals with knee OA walk with an increased inclination of the trunk?

RQ 1B: Do individuals with knee OA stand with an increased inclination of the trunk?

RQ 1C: Does trunk inclination in standing correlate with trunk inclination in walking both in a group of individuals with knee OA and also in a healthy cohort?

RQ 1D: Is there a difference in CoP between healthy and knee OA subjects?

RQ 1E: Is there a link between forward trunk inclination and anterior shift of CoP?

RQ 1F: What are the differences in hip/knee/ankle moments between healthy and knee OA subjects?

RQ 1G: What are the differences in hamstring/quadriceps/gastrocnemius muscle activity between healthy and knee OA subjects?

RQ 1H: What are the differences in the co-contraction between healthy and knee OA subjects?

Study 2: The relationship between trunk inclination and joint moments/muscular co-contraction

In the sections above, I presented a model to explain how increased trunk inclination could be linked to altered lower limb moments, muscle activation and co-contraction during human walking. Therefore, this second study sought to explore relations between specific gait characteristics and trunk inclination both in people with knee OA and also healthy control subjects. A full list of research questions is provided below.

- RQ 2A: What is the relationship between trunk inclination and hip/knee/ankle moments in people with knee OA and also in healthy control subjects?
- RQ 2B: What is the relationship between trunk inclination and hamstring/quadriceps/gastrocnemius activity in people with knee OA and also in healthy control subjects?
- RQ 2C: What is the relationship between trunk inclination and co-contraction in people with knee OA and also in healthy control subjects?

It is possible that individuals with knee OA adjust their postural strategy in response to knee pain. Specifically, patients with knee OA may adopt an increased inclination of the trunk in order to reduce knee moments. As explained above, this appears be a maladaptive strategy which may lead to increased co-contraction and therefore joint loading. Scientific evidence supports the idea that co-contraction will accelerate disease progression (Hodges et al., 2016), and therefore conservative interventions are required which can be used to reduce co-contraction. If co-contraction is associated with upper body position, then these interventions need to target trunk lean, hip joint moments or increased co-contraction. These ideas are discussed further in the next section.

2.8 Conservative management of knee osteoarthritis (OA):

Most conservative management of knee OA focuses on physiotherapy, and may include a combination of other medical approaches. Osteoarthritis does not yet have a successful treatment: instead, the goal of treatment is to manage pain. A physiotherapy programme, self-management advice, and patient education are core proposals for the management of pain in patients with knee OA, as they have short-term benefits for pain as well as for psychosocial and physical function (Deyle et al., 2000, Falconer et al., 1992). Whether these benefits are persistent however is unclear, as few studies follow patients for more than 6 months, as a result of the fact that assessment of long-term benefit demands expensive, large, complex studies. The few studies with long-term follow up have not found persistent clinical benefits and do not comprise an economic evaluation (Deyle et al., 2000, Falconer et al., 1992). Benefits have been reported with manual therapy techniques used in combination with joint mobility and strengthening exercises (Deyle et al., 2000, Falconer et al., 1992). Falconer et al. found improvements in motion (11%), pain (33%), and gait speed (11%) after 12 treatments of stretching, strengthening, and mobility exercises combined with manual therapy procedures performed in a physical therapy clinic over 4 to 6 weeks. A comparison group who received the same exercise and manual therapy interventions plus therapeutic doses of ultrasound demonstrated no additional improvement (Falconer et al., 1992). It is possible that current physiotherapy practices for managing knee OA are not effective because they do not directly target trunk lean, hip joint moment or increased co-contraction.

A number or alternatives to current physiotherapy exercise-based intervention programmes have been proposed. These include; weight management in overweight patients, assistive device prescription (Pendleton et al., 2000), bracing and footwear modifications (Kerrigan et al., 2002). In general, these approaches aim to modify the loads imposed on the knee joint by either reducing body weight or using a device to redistribute load at the knee. For example, unloader braces have been shown to be effective in alleviating pain and reducing adduction moment (Crenshaw et al., 2000). One study showed how one type of unloader knee brace significantly increased the joint space in the affected compartment in patients with arthritis (Komistek et al., 1999). However, while braces were reported to be beneficial, they may not be ideally suited for the morbidly obese or for those who have peripheral vascular disease, skin disease or an inability to apply a brace due to other physical limitations (Marks and Penton, 2004).

2.9 Footwear and insole intervention in knee osteoarthritis (OA) patients

2.9.1 Insole intervention for knee osteoarthritis

Another approach for altering knee joint loads is the lateral wedge insole. These shoe modifications are used with the aim of realigning the weight-bearing load (Fang et al., 2006) and altering the loading rate of GRF during walking (Wakeling et al., 2003). Specifically, these insoles are designed to decrease knee adduction moment, thereby slowing the progression of medial knee OA (YASUDA and SASAKI, 1987, Sasaki and Yasuda, 1987). In a recent study, Jones et al. (2014) investigated the effect of lateral wedge insoles in 70 patients suffering from osteoarthritis and confirmed that a lateral wedge decreases the external knee adduction moment in knee OA sufferers (Jones et al., 2014). This confirms the findings of other previous reports (Kerrigan et al., 2002, Sasaki and Yasuda, 1987, YASUDA and SASAKI, 1987, Yeh et al., 2014). Recently, variable-stiffness shoes with increased lateral stiffness have been tested as an alternative to lateral wedge insoles (Kean et al., 2013) and were shown to decrease peak knee adduction moment during walking in individuals with OA in comparison to constant-stiffness control shoes (Erhart-Hledik et al., 2012). However, although both lateral wedge insoles and variable stiffness shoes may alter frontal plane moments, these devices are not designed to influence trunk inclination and/or sagittal joint moments. Therefore, it is necessary to consider the potential of specially-designed footwear.

2.9.2 Footwear intervention for knee osteoarthritis

Specially designed footwear has the potential to influence the forces experienced at the knee joint and therefore to reduce knee loading patterns, potentially reducing pain. This has motivated a number of research studies which have demonstrated the possible efficacy of footwear as a clinical management strategy for knee OA (Shakoor et al., 2013, Nigg et al., 2006, Shakoor et al., 2008). Most current footwear modifications aim to modify the frontal plane loads by reducing the first peak knee adduction moment. The magnitude of this moment is determined by the magnitude of the GRF vector and the perpendicular distance from the GRF vector to the knee joint centre. By making subtle alterations in centre of pressure, the direction of the resultant GRF vector is changed and this leads to corresponding changes in the frontal plane moment arm and change in the knee adduction moment.

One example of footwear designed to alter the frontal plane knee adduction moment is that of APOS shoes (Haim et al., 2012). The concept with this type of footwear is to alter the shape of the outsole so that the CoP is shifted laterally. With this shift comes a corresponding change in the direction of the GRF which brings it closer to the centre of the knee joint and therefore reduces knee adduction moment (Haim et al., 2011). Biomechanical studies have demonstrated that APOS footwear can be used to reduce both first and second knee adduction moment in people with knee OA (Haim et al., 2012).

Other studies have looked at the effect of walking barefoot on knee adduction moments. For example, Shakoor et al. (2006) evaluated variations in gait and joint loads in patients with knee OA when walking barefoot compared with walking in modern shoes. This study reported that barefoot walking decreases knee adduction moment at the hip and knee joints, with alterations in cadence, stride, toe out angle and joint ROM observed during barefoot walking. They proposed that modern shoe designs can predispose increased joint loading in patients with knee OA, thereby influencing the progression of knee OA (Shakoor et al., 2006). In a subsequent study, Shakoor et al. (2013) compared a "mobility" shoe with patients' preferred shoe and also walking barefoot for people with knee OA. The "mobility" shoe in this study incorporated a very flexible outsole which allowed the foot to function as naturally as possible. This study reported that use of the mobility shoe for six months imitated the biomechanical effects of barefoot walking and resulted in reduced knee adduction moment.

Previous research has looked at lateral wedges and various different footwear designs for people with knee OA. In general, previously tested footwear has been designed to reduce knee adduction moments, either by influencing the position of the CoP and therefore the ground reaction force or by allowing the foot to function more naturally. However, to date, there has been minimal investigation of footwear which is specifically designed either to influence sagittal knee joint moments or trunk inclination. In the previous chapter, a biomechanical rationale was presented to explain how increased trunk inclination could affect sagittal moments and muscular co-contraction. Therefore, further research is needed into footwear which can influence sagittal loading and/or trunk inclination. One shoe design which has the potential to influence these aspects of gait is the rocker shoe.

2.10 The biomechanical effects of rocker shoes in healthy people

Rocker shoes are designed with a rigid outsole which is designed to “rock” the foot forwards during the stance phase of walking. There are a number of different types of rocker shoe (see the figures below), each of which has a different type of rocker profile. For example, the traditional rocker incorporates a distinct apex angle, whereas the toe-only curved rocker incorporates a gradual curve in the outsole geometry. Regardless of the exact profile, all rocker shoes facilitate a rolling motion of the foot as the body’s CoM moves across the profile’s fulcrum point (Myers et al., 2006). Rocker shoes have been shown to reduce plantar pressures under the forefoot (Chapman et al., 2013, Ochsmann et al., 2016, Brown et al., 2004) during walking, which is assumed to be the result of a reduction in range of motion at the metatarsal joints (Hutchins et al., 2009). However, rocker shoes also have the potential to influence sagittal joint angle, moments and upper body positioning. The evidence supporting these biomechanical changes is outlined in the sections below.

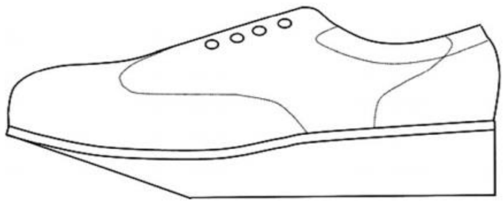


Figure 2-13. Angled rocker shoe (Traditional) (Hutchins et al., 2009)

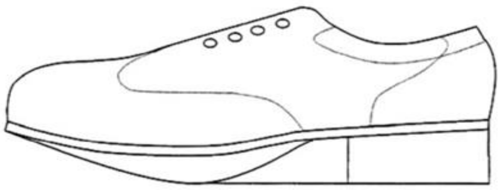


Figure 2-14. Toe-only curved rocker shoe (Hutchins et al., 2009).

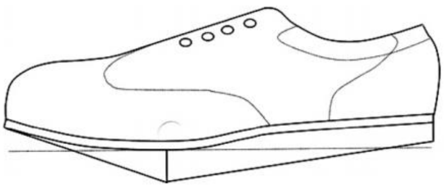


Figure 2-15. Negative heel rocker shoe (Hutchins et al., 2009).

2.10.1 Lower limb joint kinematics

Rocker shoes have been found to influence joint angles at the ankle, knee and hip. For example, studies have shown that rocker footwear can lead to a decrease in ankle plantarflexion range of moment (Gardner et al., 2014). These findings are similar to other studies of rocker footwear which shows a decrease in the knee extension angle in early stance phase and also a decreased hip extension angle in the late stance period (Taniguchi et al., 2012). However, there is some inconsistency between different studies, which may be due to the difference in the rocker profile tested. For example, MBT footwear has been shown to reduce peak hip flexion, increase peak knee extension, and reduce hip and knee range of motion throughout gait (Tan et al., 2016). In contrast, Long et al. (Long et al., 2007) tested a custom-made double rocker sole and found to increase flexion at the hip, knee, and ankle during early and mid-stance.

2.10.2 Lower limb joint kinetics

A number of previous studies have investigated the effect of rocker footwear on ankle plantar flexor moment during walking. In general, this research shows the plantar flexor and dorsiflexor moments to be reduced with rocker footwear (Taniguchi et al., 2012, Boyer and Andriacchi, 2009). However, some studies have also demonstrated alterations in knee kinetics with rocker footwear. For example, Buchecker et al. (2013) observed a reduction in the external knee flexor moment during loading response and reduced concentric hip power output during early stance. These findings contrast with the results reported by Sobhani et al. (2013), who observed no change in knee or hip moments. However, it is important to recognise that Sobhani et al. did not test shoes with the same outsole profile as Buchecker et al. (2013). Again, this highlights the importance of the precise rocker profile on the biomechanical aspects of gait which are to be modified. Importantly, the study by Buchecker et al. (2013) highlights that it may be possible to reduce hip moments during early stance with an appropriately designed rocker shoe. Given the biomechanical rationale, presented in the section above, in which increased hip moments were associated with increased trunk inclination and elevated hamstring activity, it is possible that appropriate rocker footwear may be effective at reducing hamstring activity and therefore joint loading in people with knee OA.

2.10.3 Muscle activity during walking

By using EMG techniques, a number of previous studies have sought to understand whether muscle activity during walking is altered when wearing rocker footwear. Again, the findings of these studies differ depending the precise design of the rocker outsole profile. For example some studies have shown increases in gastrocnemius activity (Forghany et al., 2014). In contrast, Sobhani et al. (Sobhani et al., 2013) showed a delayed onset of the triceps surae muscle group when walking with rocker shoes and Santo et al. (Santo et al., 2012) found no change in muscle activity in either biceps femoris, gastrocnemius or tibialis anterior when walking in rocker footwear. To date, most studies have demonstrated minimal changes in the EMG activity of the other knee extensors/flexors. However, it is still not clear whether, with an appropriately designed shoe, it would be possible to reduce hamstring activation.

2.10.4 Trunk motion during walking

Very few studies have investigated the effects of rocker footwear on trunk position. While one study showed that rocker footwear might increase trunk flexion (Talaty et al., 2016), another showed a clear decrease in trunk inclination when wearing rocker shoes (Ochsmann et al., 2016). Again, the different types of rocker shoes tested were different between these studies. However, the results of the latter study illustrate the possibility that appropriately designed rocker footwear could be used to decrease trunk inclination in people with knee OA.

Rocker footwear is designed based on the principle of introducing instability in order to influence parameters of posture and gait. Performing any given task involves interactive effects from the simultaneous coordination of many muscles and joints (Sousa & Tavares, 2014). As an individual moves, the central nervous system receives a range of inputs, which it must prioritise and respond to in order to maintain balance, by adjusting the way the body is aligned.

It is possible that the introduction of instability (such as a rocker shoe) may accentuate the effect of poor upper body position and this might make the individual feel more imbalanced. In order to adapt to this potential instability, they will be required to improve overall segmental alignment and therefore balance. This leads to the idea that introducing instability at the foot could lead to adjustments to posture through the central nervous system (Sousa & Tavares, 2014).

As explained above, previous studies have demonstrated that rocker footwear can influence trunk inclination (Ochsmann et al., 2016, Sousa et al., 2014, Talaty et al., 2016), with one study showing a decrease in trunk inclination in healthy people (Ochsmann et al., 2016). This research indicates that rocker footwear could have a potentially positive effect on postural control performance. Interestingly, other research has demonstrated lower latency in the ankle muscles, decreased gastrocnemius activity, as well as changes in the activity of the thigh muscles both immediately and after long term use of unstable footwear (Sousa & Tavares, 2014). Taken together, these data suggest that rocker footwear can alter both muscle activity and postural control and suggest that the destabilising effect of the rocker shoe could encourage a realignment of body segments, with associated changes in muscle activity. This idea is explored in the final chapter of the thesis.

2.10.5 Biomechanical and clinical effects of rocker shoes in people with knee OA

To date, only three studies have investigated either the biomechanical or clinical efficacy of the rocker shoe for individuals with knee osteoarthritis (Nigg et al., 2006, Tateuchi et al., 2014b, Madden et al., 2015). Nigg et al.'s (2006) study was a randomised controlled trial with 123 subjects and focused on one type of rocker shoe, known as the Masai Barefoot Technology

(MBT) shoe. Although this study did not include gait-related biomechanical outcomes, the authors did look at static and dynamic balance and knee joint range of movement measures, in addition to clinical outcomes investigating pain and function. The study demonstrated improvement in static balance for the MBT group and which were accompanied by reductions in pain. The authors concluded that rocker shoes may be effective in reducing pain in people with knee OA.

A study by Tateuchi et al. (2014) focused on the immediate biomechanical effects of MBT shoes in women with knee OA. This study showed that MBT rocker shoes decreased the knee flexion moment in early stance but observed no change in the peak knee extensor moment. However, they did report an "increase in trunk lean toward the extension direction while wearing the MBT shoes." This suggests that the MBT rocker shoes actually reduced forward lean in their OA patients and this may explain their observed reduced in knee flexion moment. However. Tateuchi et al. (2014) did not report hip extension moments or EMG activity and therefore it is unclear whether the change in trunk inclination was accompanied by a corresponding change in the hip extensor moment.

In a more recent study, Madden et al. (Madden et al., 2015) investigated the immediate biomechanical effects of another type of rocker shoe design in people with knee OA. Specifically, they focused on the Sketchers Shape-ups, a commercially available rocker shoe. Their data showed a significant decrease in the peak knee adduction moment with the rocker footwear but no corresponding change in the knee flexion or extension moment. Again, the difference between these findings and those of Tateuchi et al. (2014) highlight the different effects of different rocker shoes and demonstrate that, if rocker footwear is to be effective for reducing knee joint loading in knee OA, there needs to be careful selection of the outsole profile.

2.10.6 The three-curved rocker shoe (T3C)

It is possible that a very specific design of rocker shoe is required to alter trunk inclination, reduce hip moments and reduce knee muscle co-contraction in people with knee OA. In a recent study, Hutchins et al. (2012) proposed a rocker shoe design with a rocker profile created from three different curves, generated from circles drawn around the hip, knee and ankle. This design aims to maintain a normal walking pattern but to reduce muscle activation around the hip, knee and ankle, and therefore may be an ideal candidate to test in a cohort of patients with knee osteoarthritis. The shoe design contains three curves positioned so that during the stance phase of walking, the radius of each curve is centred on the sagittal plane centres of three lower limb joints; ankle, hip and knee, in that order, as shown in Figure 2-15. This specific design helps the shoe to roll forward gently and it has been suggested that it may reduce muscle activity around the knee and ankle (Hutchins et al., 2012). It is possible that any reduction in the hip extensor moment may be accompanied by a corresponding reduction in hamstring activity and this may lead to a reduction in co-contraction of the hamstring-quadriceps muscles.

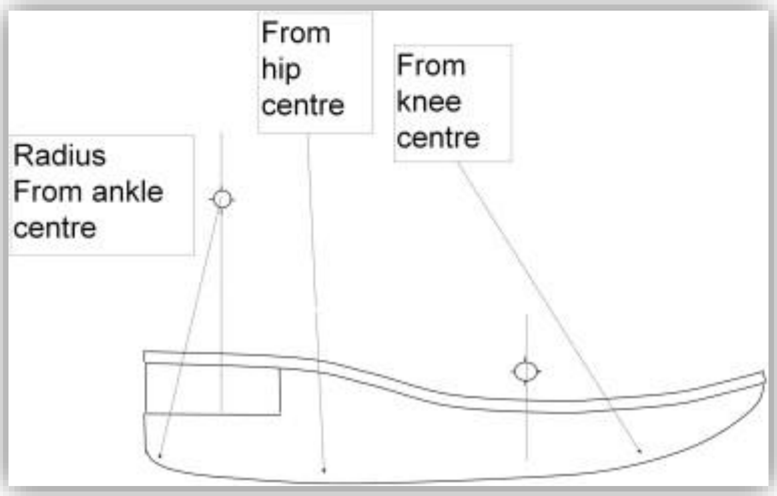


Figure 2-16. Radiuses of 3 curve rocker shoe (T3C)

Hutchins et al. (2012) suggest that the three-curve rocker shoe could alter the direction and orientation of the GRF during walking in such a way as to reduce the ankle joint moment during early stance. Such a change in ankle moment may lead to a corresponding reduction in the activation of the gastrocnemius muscle during walking and it is possible that this would lead to a corresponding decrease in gastrocnemius-quadriceps muscle co-contraction during mid-stance. Given the potential for the three-curve rocker profile to influence both hip moments and ankle moments and therefore muscle co-contraction, further investigation is required to establish the precise biomechanical effect of this footwear design in a cohort of patients with knee OA.

2.10.7 Summary and research questions

The gait of people with knee OA is characterised by increased hip extension moments, decreased knee extension moments and increased muscular co-contraction during 15-25% of stance phase. Previous research has demonstrated that rocker footwear has the potential to alter trunk inclination, lower limb moments and also muscle activation patterns during normal walking. However, these effects appear to be strongly dependent on the exact rocker profile incorporated into the footwear. The three-curve rocker profile has been designed to reduce both hip and the ankle moment in early-mid stance and therefore may be effective at reducing co-contraction during 15-25% of stance phase. Therefore, in the final study of this thesis, the following questions will be addressed:

Study 3: The biomechanical effects of rocker footwear in people with knee OA

- RQ 3A: How does trunk inclination change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?
- RQ 3B: How do lower limb moments change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?
- RQ 3C: How the activation patterns of the specific muscular
- RQ 3D: How does muscular co-contraction change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?
- RQ 3E: Are there immediate changes in the measurement of pain when people with knee OA wear a three-curve rocker shoe?

2.11 Overarching aim of the thesis

In the section 2.5-2.7, I proposed a new framework which may explain some of the previously observed differences between people with knee OA and heathy controls. This framework was based on the idea that an increase in forward trunk inclination could alter the direction of the ground reaction force vector and therefore the moments and muscle activation patterns at the hip, knee and ankle. In section 2.8-2.10, I discussed the idea that, with a simple footwear intervention, it may be possible to influence upper body position and/or change the direction of the ground reaction force during walking. Together these different ideas motivate the overarching aims of the thesis. These aims are to characterise forward trunk inclination in people with knee OA (Chapter 4), understand whether there is a link between trunk inclination and specific biomechanical variables related to loading (Chapter 5) and understand whether footwear-based intervention could bring about changes in trunk inclination and/or biomechanical variables related to joint loading (Chapter 6).

Chapter 3 - Methodology

3.1 Overview of the data collection procedures

A single data collection session, for each participant, was used to obtain the necessary data to address the overarching objectives described above. Although these data was collected at the same point, it was analysed separately in order to address the three separate objectives of the thesis, subsequently referred to as studies. The first study examined the differences between individuals with knee OA and healthy participants in terms of the forward trunk inclination during walking and standing, gait characteristics such as anterior-posterior displacement of centre of pressure (A-P CoP) and sagittal moments of hip, knee and ankle. Additionally, the differences in muscle activity between people with knee OA and healthy individuals were investigated for the hamstring (biceps femoris and semitendinosus), quadriceps (vastus medialis (VM) and vastus lateralis (VL) and gastrocnemius (medial and lateral gastrocnemius). as well as muscle co-contraction.

In the second study, the link between forward trunk inclination during walking and sagittal moments of lower limb joints hip, knee and ankle were assessed for both subject groups. In addition, the link between forward inclination and both muscle activity and muscle co-contraction was also assessed, again for both groups. In the final, third study, the biomechanical effect of the three-curve rocker shoe on forward trunk inclination, sagittal lower limb joint moment, muscle activity and co-contraction was investigated for both healthy and OA subjects.

As explained above, a single protocol for data collection was developed to address each of the three studies, as shown in Figure 3-1, and described in detail in the sections below. Although these data were collected during the same session, extensive analysis and processing was performed on each dataset and specific outcomes derived/ statistical analysis performed for each of the separate studies. These outcomes and the statistical analysis are described in the subsequent chapters. It was deemed appropriate to present the data in three separate studies (corresponding to chapter 4,5 & 6) in order to ensure clarity of the thesis. Furthermore, although it would have been possible to collect the data for the final footwear study through second testing session, this would have been more inconvenient for the participants. Therefore, the decision was made to collect all data in a single longer session, which incorporated a break midway through. In total, data were collected on a total of 47 subjects; a cohort of 27 knee OA participants as well as 20 healthy participants.

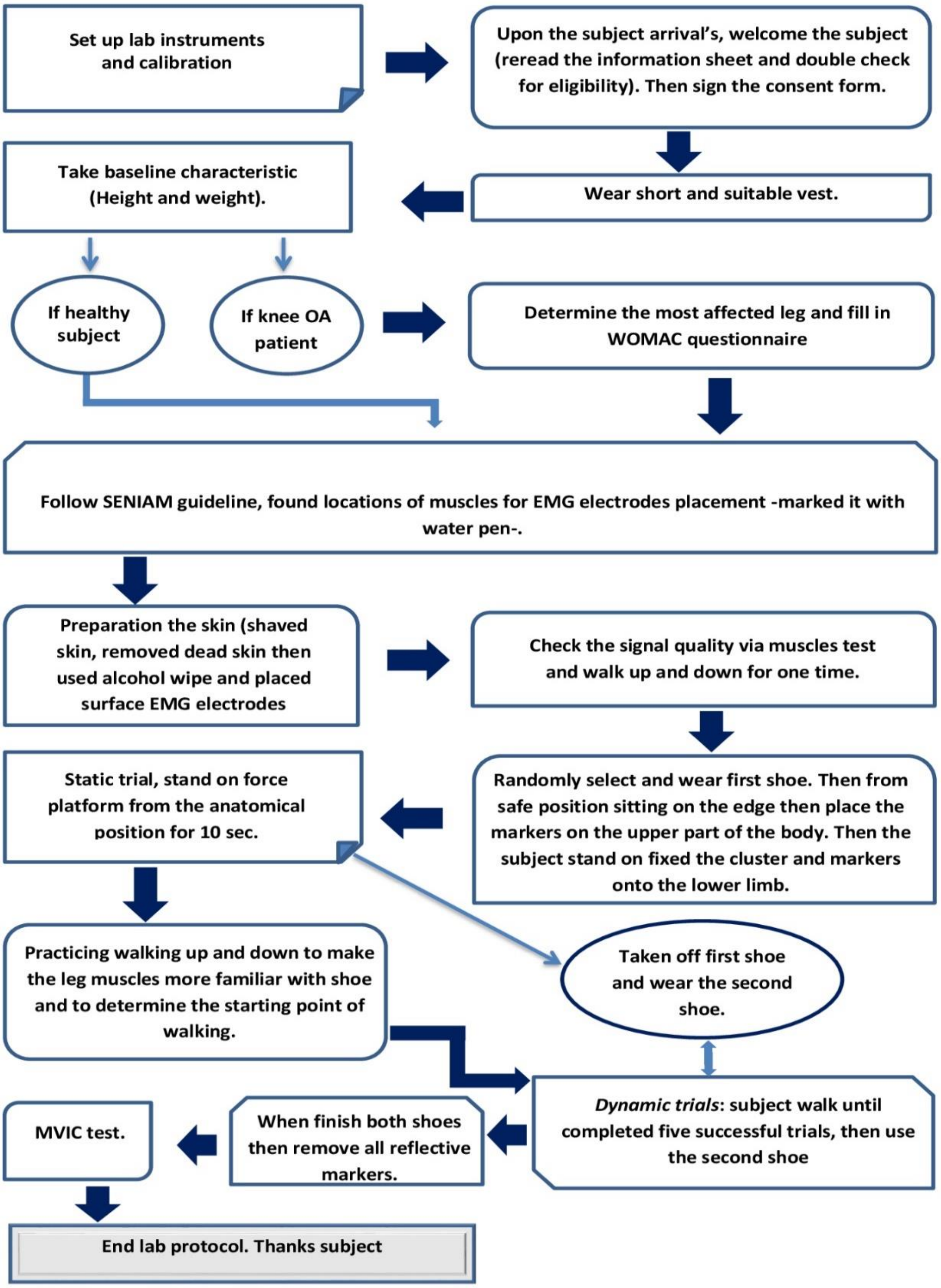


Figure 3-1. Steps of lab protocol

3.2 Ethical approval, participant recruitment & sample size calculation

The University of Salford staff and PGR ethics committee provided ethical approval for the study (Reference HSCR15-35). Approval was also obtained from the NHS ethics committee via the IRAS system (Reference 15/EM/0502). Letters of approval for each of these panels, along with the participant information sheet and consent forms, are provided in Appendices.

3.2.1 Recruitment strategies:

Recruitment of knee OA and healthy participants was done through a range of approaches. Posters (flyers) and emails were sent around on the campus of the university to both students and staff. The emails and flyers contained the lead researcher's contact details and potential participants were required to contact the lead researchers directly if they were interested in volunteering.

Another avenue for recruitment was the Citizen Scientist Salford Project. This project represents a collaboration between Salford city council, GM universities and Salford NHS trust and recruits individuals to participate in research projects. This is achieved through listing projects on a dedicated website, as well as via a monthly newsletter. In addition to the citizen scientist website, the School of Heath Sciences website was used in a similar way to advertise this project (http://www.salford.ac.uk/health-sciences/research/opportunities-to-participate) Again, anyone interested in participating in the study was required to contact the lead researcher directly.

The final avenue for recruitment was via a volunteer database held at the University of Salford. A database search was carried out by the manager of the database, and potentially eligible people send an invitation letter and participant information sheet. Again, those interested in taking part were required to contact the lead researcher directly.

The recruitment approach described above allowed identification of both healthy and knee OA subjects, leaving control over contacting the researcher and volunteering for the study with the potential participants. After contact had been made, volunteers were called or emailed and asked a number of questions to identify whether or not they were eligible according to the exclusion and inclusion criteria. Those who qualified were sent the participant information sheet through post or email and allowed a minimum of three days to read and absorb the information contained on the sheet, before being re-contacted and given an invitation to attend the test at the university. Every participant who took part in the study was pad £25 in cash to cover travel expenses and give compensation for the loss of 2 hours of their time required to participate in the research project.

3.2.2 Sample size calculation

In order to establish an appropriate group size for each of the studies in the thesis, power size was calculated a priori using g-power software following Faul, Erdfelder, Buchner and Lang (2007; 2009). The calculations made are explained below for each of the main research studies:

Study 1 (Chapter 4): A sample size of 20 individuals was chosen for this research question, which looked for differences between healthy and knee OA individuals across a range of parameters. The primary outcome in this study was trunk inclination during walking, and based on previous work by Turcot et al. (2015), I estimated an effect size of 1 standard deviation as a difference between the two groups (healthy and knee OA). The study by Turcot et al. (2015) investigated standing joint angles for the trunk, pelvis and hip with a mean ± SD (OA: 5.3°± 8.7°; control:−3.5° ± 5.6°). This was taken as the best indicator for differences between the two groups which would be associated with walking. With $\alpha = 0.05$ and a power of 0.8 based on using a two-tailed test, the required minimum sample size for the study is n=17 in each separate group.

Study 2 (Chapter 5): The primary focus of this study was the correlation between trunk inclination in walking and moments. No similar investigation of this type of relationship has been carried out before and so it was difficult to estimate the potential correlation. Therefore, I based my sample size calculation on a correlation of r=0.5, which represents a weak-moderate correlation. Based on this correlation coefficient, with an alpha of 0.5 and a power size of 0.8, the g-power software (Faul et al., 2007; Faul et al., 2009) estimated that I would need a sample size of n=26.

Study 3 (Chapter 6): This primary objective of this study was to investigate the change in biomechanical parameters between the different footwear conditions (control and rocker shoe). Previous studies have reported different effects, on trunk lean, of rocker footwear between 0.35 and 1.25SD. Specifically (Tateuchi et al., 2014a) observed a mean (SD) of −0.1° (4.7°) for control shoes and a mean (SD) −1.6° (4.3°) for MBT rocker shoes, whereas Ochsmann et al. (2016) observed a mean (SD) of 8.9° (2.2°) for control shoes and a mean(SD) of 5.9° (2.4°) for rocker shoes. Given these different effects, I based the sample size calculation around a conservative estimate of a 0.5SD difference between the control and the rocker shoe. Using the g-power software, with a power = 0.8 and $\alpha = 0.05$, it was shown that a sample of n=25 would be needed to identify a 0.5 SD difference between footwear conditions.

3.3 Inclusion/exclusion criteria:

The following inclusion/exclusion criteria were used for the participants with knee OA and also the healthy participants.

Inclusion criteria (both groups):

- Age range of 40-85 (upper age limit due to the amount of walking involved in the study).
- Ability to stand and walk independently.
- Ability to speak and understand written English
- Ability to walk without any walking assistance for at least 250 m.

Exclusion criteria (both groups): Participants were excluded if they satisfied any one of the following exclusion criteria:

- Complex pain conditions such as diabetic neuropathic pain, fibromyalgia
- Previous surgery to the lower limb.
- BMI >33, since it is not possible to perform accurate measurements on individuals with excess adipose tissue.
- Lower limb arthroplasty.
- Any systemic inflammatory disorders, such as rheumatoid arthritis.
- Any balance disorders which may increase the risk of a fall.
- Low back pain.

Participants with knee OA were defined using the following criteria:

- Clinical diagnosis of knee OA affecting the tibiofemoral joint, according to American College of Rheumatology (ACR) guidelines (Altman et al., 1986).

- Patients with patellofemoral OA were excluded. Such patients were identified as reporting pain or discomfort within the patellofemoral joint.
- Pain for at least 6 months' duration (if they are a participant with knee OA).
- Pain or difficulty in rising from sitting and/or climbing stairs (if they are a participant with knee OA).

3.4 Participant characteristics

Recruitment for this project was challenging, for a number of reasons. Firstly, there were a number of other knee OA research projects happening at the same time as this project and this put pressure on the people registered on the database, many of whom did not have the time to participate in multiple projects. Another challenge to recruitment was the need to identify people with knee OA who had a BMI lower than 33. This was necessary to obtain high-fidelity EMG signals, but excluded many people with this disease. Finally, some potentially eligible participants decided not to participate in this project due to the complexity of the gait lab protocol. Nevertheless, after persisting with recruitment, I was able to identify and test at total of 47 participants. The demographics of the 47 are detailed in Table 3-1 (knee OA) individuals and Table 3-2 (healthy participants). It can be seen that, in the participants with tibiofemoral OA 13 had a more affected right side and 14 a more affected left side. Importantly, both health and OA groups were similar (no significant differences) in age, height, weight and BMI. With the protocol we asked the healthy people to walk slightly slower than normal (see later) and this resulted in similar walking speeds between the two groups. Table 3-3 shows no significant difference between the groups relating to age, height, weight, and walking speed.

Table 3-1. Knee OA demographic

Items		*People with knee osteoarthritis OA (tibiofemoral) (n = 27)*	
		Mean (SD)	*Range (maximum-minimum)*
Age (yrs)		*56.71(± 8.8)*	*35*
Weight (in kg)		*82.08 (±11.06)*	*35*
BMI (kg/m^2)		*27.74 (±3.4)*	*11.7*
Height (m)		*1.72 (±0.06)*	*.28*
Mean walking speed (in M/s)		*1.07(± 0.13)*	*0.80 -1.3*
WOMAC	*pain Mean (SD)*	***9.41 (±**2.62)*	
	Function Mean (SD)	***4.37 (±**1.75)*	
	Stiffness	***28.41 (±**10.05)*	
Affected side		*13 R 14 L*	

Table 3-2. Healthy participants demographic

Items	*Healthy participants (n = 20)*	
	Mean (SD)	*Range (maximum-minimum)*
Age (yrs)	*53.65(± 11.3)*	*33*
Weight (in kg)	*81.2 (±11.8)*	*44*
BMI (kg/m^2)	*27.13(±3.1)*	*7.2*
Height (in m)	*1.73 (±0.07)*	*0.3*
Mean walking speed (in M/s)	*1.09(± 0.13)*	*.44*
Dominant side	*2L 18R*	

Table 3-3. Independent t-tests showed that there was no difference in age, weight, height between the two groups

Items	Mean (SD)		***P***
	Knee OA	Healthy participants	
Age (yrs)	57 (9)	54 (11)	0.28
Weight (in kg)	82 (11)	80 (12)	0.51
Height (in m)	1.7 (0.07)	1.7 (0.07)	0.52
Mean walking speed (in M/s)	1.1 (0.13)	1.09 (.013)	0.81

3.5 Clinical characterisation of knee OA and overview of the data collection procedure

Upon the subjects' arrival at the gait laboratory at Salford University, participants were provided with a brief introduction to the gait laboratory equipment, after which the study aims were explained to them to ensure that they were happy to participate. The investigator then asked participants to reread the participant information sheet and to sign a consent form to take part in the study. The knee osteoarthritis subjects then completed the Western Ontario and McMaster Universities Osteoarthritis (WOMAC) clinical questionnaire, which is used to capture pain, stiffness and function in patients with knee OA. This is a standard questionnaire which is used regularly in clinical trials of knee OA (Roos et al., 1999, Theiler et al., 2002).

Knee OA was characterized using the ACR criteria for clinical classification, which are widely used for this condition (Altman et al., 1986). For knee OA classification, the ACR criteria require pain to be present in the knee as well as 50% or more of the 6 characteristics given below (Altman et al., 1968):

- Age > 50 years old
- Morning stiffness < 30 minutes
- Crepitus on knee motion
- Bony tenderness
- Bony enlargement
- No palpable warmth

All participants recruited into this study had received a previous diagnosis of knee OA from a qualified medical practitioner, such as a GP. Patients for whom the patellofemoral joint was affected were excluded to ensure that only subjects with OA in the tibiofemoral joint were studied. I chose to rely on ACR criteria to confirm the presence of knee OA because, although some of the participants had had their diagnosis confirmed via x-ray, it was not possible to access these x-rays. Furthermore, it was not deemed ethical to send those who had not been x-rayed for a scan because of the unnecessary exposure to ionising radiation. This approach, of identifying knee OA based on the ACR criteria, is supported by previous research by Skou et al. (2014), which shows that x-rays are unnecessary for diagnosis of knee OA and recommends that radiological assessment is only needed when further investigation is required to investigate other pathological conditions (Skou et al., 2014). As the aim of this thesis was not to examine radiological progression in knee OA, it was deemed appropriate to use ACR criteria as a diagnostic tool. This is in line with current clinical practice, advocated by the National Institute for Health and Care Excellence (NICE), quality standard QS87 (2015).

Following consent and recording of the ACR criteria, the investigator recorded demographic details such as age, weight (kg) and height (m) and determined the correct shoe size for the individual. Participants were then asked to change into their shorts and a comfortable vest in a private room. The full experimental procedure is set out in the diagram below and listed in Appendix 1. Full details of each procedure are provided in the sections which follow.

3.5.1 Clinical outcomes

The WOMAC (Western Ontario and McMaster Universities) index for osteoarthritis was used to rate each participant in the study, prior to gathering biomechanical data. Each question was answered based on a 5-point Likert-type rating system with were 5 questions for pain, 2 for stiffness, and 17 for function. The WOMAC score is the questionnaire most often utilized in existing knee OA research (Bellamy, Buchanan, Goldsmith, Campbell, & Stitt, 1988). Further, it is accepted as sufficiently valid, responsive and reliable (Chesworth, Mahomed, Bourne, & Davis, 2008; Escobar et al., 2007; Jinks, Jordan, & Croft, 2002; Ryser, Wright, Aeschlimann, Mariacher-Gehler, & Stucki, 1999; Salaffi, Carotti, & Grassi, 2005). Each participant also filled in the health history questionnaire in order to provide information on their existing and previous medical treatment and conditions.

3.6 EMG measurement procedures

3.6.1 Background to EMG measurement

Electromyography (EMG) is used to record and evaluate the electrical activity which skeletal muscle generates when muscles contract (Reaz et al. 2006). The term motor unit refers to the lowest size of functional unit which is a group of muscle fibres all innovated (and therefore activated) by the same motor neuron. The term MUAP (motor unit action potential) refers to an action potential which is generated when the motor unit becomes active and when specific biochemical changes occur in the motor unit. This action potential propagates along the motor unit and can be detected using invasive (in-dwelling wires) or using surface EMG techniques. Surface EMG techniques involve the use of surface electrodes which are typically 1-2cm apart which record the summed activity of many motor units. This signal can be analysed to provide information on activity levels and temporal characteristics of muscle activation (Preece et al., 2016).

There are two different types of EMG collection systems can be used to record the muscle activity signal: surface electromyography and fine wire electromyography. The first approach is widely used in measuring muscles' activity while walking, due to its non-invasive nature and high level of safety (Kleissen, Buurke, Harlaar, & Zilvold, 1998). However, the typical size of signal in walking is (100-500 microvolts) while in rest typically is about 5-10 microvolts. The magnitude of the surface EMG signals is influenced by the type of tissue present (e.g. subcutaneous fat), as well as how thick the tissues are (Farina & Mesin, 2005), the ambient temperature and physiologically-based crosstalk (Winter, Fuglevand, & Archer, 1994), as well as altered geometrical relationships between the location of the electrodes and the muscles being measured (Delaney, Worsley, Warner, Taylor, & Stokes, 2010). Furthermore, surface EMG is negatively impacted by noise from outside sources. In comparison, fine wire EMG is based on inserting electrode needles into the muscles themselves, which can cause discomfort or pain (Perry, 2010). Moreover, this approach does not detect the signal from the whole muscle but only from the motor unit where it is inserted, and the pain caused can have an impact on the way the muscle is activated. For these reasons, surface EMG is typically used to analyse the activation of muscles in during gait tasks, especially for pathological groups.

3.6.2 EMG data collection system

Collection of data from surface electromyography (EMG) was carried out via a Noraxon Desktop Direct Transmission System (DTS), sampling at 1500 Hz, as shown in Figure 3-3 The system was configured with eight channels for EMG use, and was used in conjunction with Qualisys motion capture system in the laboratory while biomechanical tests took place. Equipment needed for measurement of muscle activity via DTS in the study included sensors, chargers, sensor cables/leads, USB leads, DTS Wireless EMG Sensors, double-sided tape, dual electrodes (figures3-2 to 3-5), the myomuscle program and desktop USB receiver. From the eight channels of the EMG DTS, six were employed, in line with the requirements for muscles to be monitored along with single-use adhesive Ag/AgCl snap EMG electrodes shaped in a figure of eight, and measuring 2.2x4cm, with two 1 cm-diameter conductive circles and 2 cm separating each electrode. The electrode equipment comes with hypoallergenic adhesive and gel and similar to accepted practice Ag/AgCl electrodes were used (Hermens et al., 2000). With the DTS system, information from the muscle being monitored is transmitted, via Bluetooth, between the sensor electrode and the receiver on the desktop. This allows EMG measurements to be taken simply, as it bypasses the requirement to connect the EMG base station and the electrode (subject) with a connecting cable.

Figure 3-2. Desktop DTS

Figure 3-3. SEMG sensor

Figure 3-4. EMG lead set 3 inches electrode cable

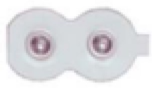


Figure 3-5. Dual electrodes

Figure 3-6. Nuprep gel.

3.6.3 Skin preparation, electrode placement & signal testing

3.6.3.1 Skin preparation

Before EMG measurement, the skin must be prepared in order to reduce the impedance of the skin and enable the electrode to be firmly attached, improving the quality of the signal. Prior to recording EMG signals, if required an area of 2-3cm across was shaved with a disposable razor (if appropriate). After that, Nuprep Gel (Figure 3-6) was used as an exfoliant at the location identified for electrode placement, in order to remove dead skin. A wipe with 70% isopropyl alcohol was used for cleaning of the area, before leaving the skin for 2 minutes to dry. Finally, electrodes (disposable adhesive Ag/AgCl snap EMG electrodes with 20 mm between each electrode) were placed on the skin oriented parallel to the direction of muscle fibres. Preparing the skin and placing the electrodes appropriately are both important measures to record EMG signals of good quality.

Electrode placement & signal testing

The first part of the experimental protocol was to place EMG electrodes over six muscles on the most affected side if bilateral knee OA was present, or on the affected leg if unilateral (or matched leg in healthy group): VM, VL, semitendinosus, biceps femoris, medial and lateral gastrocnemius. There are various guidelines available for placement of electrodes, but the most frequently used is the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) Guidelines (http://www.seniam.org). SENIAM guidelines were implemented in this study to select electrode sites through locating muscles using given reference points in the anatomy. The SENIAM guidelines offer a method for minimising the risk of crosstalk in the signal from the muscles which surround the target muscle. Further, it is recommended in the SENIAM guidance that muscles are palpated as the subject performs manual resisted isometric contractions, allowing the researcher to locate the muscle belly and reference points in the anatomy (Hermens et al., 1999), minimising crosstalk occurrence (Winter, Fuglevand & Archer, 1994). Positioning participants in an initial position which is specified by SENIAM for individual muscles allows electrodes to be correctly positioned (Hermens et al., 1999, Winter et al., 1994). The muscle belly was palpated to establish muscle layout and fixed features in the anatomy to locate sensors correctly. The muscle palpations were recommended to establish the underlying muscle layout (Hubley-Kozey et al. 1998; Leveau et al.1992). Using the procedure described above, the 6 muscles tested were located in line with SENIAM guidance, and as described below:

Biceps femoris (lateral hamstring)

The participant was asked to lie face down on the bed and the thigh was kept on the bed while the knee was flexed to an angle smaller than 90°, while laterally rotated. For testing, participants were asked to try to flex the knee against a fixed resistance (the experimenter held the foot of the participant). This allowed the muscle belly to be palpated. Placement of the electrode was made at a distance of halfway from the ischial tuberosity to the tibial lateral epicondyle and over the belly of the muscle.

Semitendinosus (medial hamstring)

The same procedure as described above was followed, however, the leg was medially rotated. For this muscle, the electrode was placed halfway from the ischial tuberosity to the tibial medial epicondyle over the muscle belly.

Vastus medialis (medial quadriceps)

To locate electrodes on the vastus medialis, participants sat on the bed with their back supported in a slight backward inclination of the upper body. Testing was carried out with the leg straight. Electrodes were positioned at 80% of the way from the anterior superior iliac spina (ASIS) to the joint space found before the medial ligament's anterior edge. Testing to confirm the muscle belly had been correctly located was carried out by participants extending their knees with as the examiner resisted any movement of the leg by holding the ankle.

Vastus lateralis (lateral quadriceps)

Participants were positioned as described above and electrodes placed at 66% of the distance between the anterior superior iliac spina (ASIS) and the patella at its lateral side. Testing that the location of the muscle belly had been identified was carried out by participants extending their knees as the examiner applied pressure at the ankle to resist leg movement.

Medial gastrocnemius

Electrodes for the medial gastrocnemius were located with the participants lying prone on the plinth with a straight leg and the foot positioned past the end of the bed. Electrode placement was at the most prominent bulge of the muscle. Testing to confirm the location over the muscle belly was carried out by again asking participants to plantar flexion their ankle against manual resistance.

Lateral gastrocnemius

For the lateral gastrocnemius, electrodes were positioned two thirds of the way from the heel to the head of the fibula as the subject lay prone with their foot was off the plinth edge. Electrode placement was at third of the line between the head of the fibula to the heel. The protocol to check the signal quality was the same as for the medial gastrocnemius, which is explained above.

After applying the protocol, described above, for the individual muscles, the subject was required to walk up and down the gait lab as the researcher visually inspected the signals during walking to ensure that there was no signal artefact due to motion of the cable. No EMG signals were recorded within ten minutes of placement of the disposable electrodes (Redfern et al., 1994).

3.6.4 Reference contractions (types and rationale)

EMG signal magnitude is dependent on the level of muscle activation but also on a range of physiological factors, including skin conductivity, thickness of subcutaneous fat and number of motor units in the EMG collection volume (De Luca, 1997). Therefore, normalisation of EMG signals is essential if comparisons are to be made between different individuals. Normalisation is done by division of the signal by an appropriate reference contraction value (Burden, 2010; De Luca, 1997) and enables EMG signal to be shown as a percentage of reference values, meaning that different muscles and participants can be compared (French, Huang, Cummiskey, Meldrum, & Malone, 2015). A range of different normalisation methods have been proposed previously which include, maximum voluntary isometric contractions

(MVIC), sub-maximal voluntary isometric contraction, peak dynamic during gait and mean dynamic methods.

With the peak dynamic approach, processed EMG signals are normalised using the peak activation of the muscle over the movement of interest, such as walking. This results in a signal, which varies between 0 and 1 (corresponding to the peak activation). In contrast, normalisation to an MVIC involves collection of additional EMG data during a maximal reference contraction, in which the participant exerts a maximal effort. This maximal contraction is then used to normalise the movement EMG signal. Although this approach can be problematic, especially for people with pathology, it is widely used to examine neuromuscular changes in walking for knee OA subjects (Lewek et al, 2004; Hubley-Kozey et al., 2006; Ramsey et al., 2007b; Rudolph et al., 2007) and is the most frequently used method of EMG signal normalization (MIRKA, 1991). This is because of the relative disadvantages of using the peak or the mean dynamic peak normalization which do not allow quantification of the degree of overall muscle activation. This is essential when investigating people with knee osteoarthritis who will often activate their muscles much more than healthy participants. Given these limitations of the peak and mean dynamic method, I chose to use MVIC to normalise the EMG signals in this Ph.D.

In order to use MVIC normalisation techniques, it is necessary to select appropriate positions, for each individual muscle, in which to collect the normalisation data. Although there have been a large number of previous studies which have used MVIC methods to normalise EMG signals in knee OA research, there is little consistency between these studies, with different investigators using a range of different procedures for positioning the lower limb during the MVIC tests. For example, Preece et al. (2016). obtained the MVIC from the hamstrings and quadriceps muscles with the knee flexed at 45°, whereas other investigators have tested people in a position with the hip and knee in 90° degrees (Lewek et al., 2004b) or a position of 90° degree hip flexion and 70° degrees knee flexion (Winters and Rudolph, 2014). The aims of any MVIC test should be to elicit a maximal contraction from the tested muscle. However, it is also important that the same limb positioning is used for each subject. Therefore, in order to identify the optimal limb position (i.e. that which would elicit the highest muscle contraction), I performed a pilot study, which is described below.

3.6.5 Pilot Study to determine the most appropriate reference contraction

Aim:

The purpose of this pilot study was to investigate the different methods of normalisation with the aim of identifying which method was able to elicit maximal muscle contraction in each of the separate muscles (lateral gastrocnemius (LG), medial gastrocnemius (MG), vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF) and semitendinosus (ST).

Method:

The sample of this pilot study consisted of 12 healthy subjects and 16 individuals with knee OA aged between 40-85years, each of whom were tested in a range of different MVIC positions. The MVIC protocol included 10 contractions in a number of different positions (described below). Data were recorded from all muscle during all contractions, following practice contractions for familiarization. Each of the MVICs lasted for 3 seconds, followed by a one-minute rest period (Hubley-Kozey et al., 2006). The tests for the three muscle groups are described below:

Hamstrings:

For biceps femoris and semitendenosis muscles, the participants lay on the bed and were then instructed to push their heel into the researcher’s hand (flexing their knee). Resistance was applied to the participant’s movement manually while the researcher’s hand remained in a consistent position. The task was performed in different positions and a goniometer was used to determine the angles of the leg (Rutherford et al., 2011). The following positions were tested:

- Resisted knee flexion at 15 degrees and leg rotated slightly inwards. (KF15 IR)
- Resisted knee flexion at 15 degrees and leg rotated slightly outwards. (KF15 ER)
- Resisted knee flexion at 55 degrees and leg rotated slightly inwards. (KF55 IR)
- Resisted knee flexion at 55 degrees and leg rotated slightly outwards. (KF55 ER)

Quadriceps

For VM and VL muscles, the participant sat on a chair and was instructed to extend their knee against the bar (figure 3-7) (which was padded to minimise discomfort). The task was performed in different four positions and a goniometer was used to determine the angles. The four testing positions were:

- Resisted knee extension at 15 degrees and leg rotated slightly inwards. (KE15 IR)
- Resisted knee extension at 15 degrees and leg rotated slightly outwards. (KE15 ER)
- Resisted knee extension at 45 degrees and leg rotated slightly inwards. (KE45 IR)
- Resisted knee extension at 45 degrees and leg rotated slightly outwards. (KE45 ER)

Gastrocnemius:

For gastrocnemius muscles, the participants first sat on the plinth with their back supported and were instructed to plantarflex their ankle against the bar (PF sit). Specifically, they were asked to push against the ball of their foot and a strap was used to prevent knee flexion. A second task for gastrocnemius muscles involved the participant standing on one leg. In this position, they were instructed to plantarflex their ankle against body weight (as forcefully as they could) while holding the frame for balance only (PFstand).

Figure 3-7. Movable bar for MVIC

With the testing described above, there was a total of 10 different MVIC tests carried out. Each was performed once and for each test, EMG data was collected using the Noraxon system for the six different muscles. Note that this testing was always performed after the gait analysis testing to avoid any possible effects of fatigue on the gait assessment.

Following data collection, MVIC data was exported to a custom Matlab programme which was used to carrying out specific signal processing. Firstly, a 20Hz high pass FFT filter was used to remove noise and movement artefact. The signal was then rectified and a low pass 6Hz Butterworth filter used to create a linear envelope. This this linear envelope frequency has been used in a number of previous studies (Winter 1990; Hubley-Kozey et al. 2006). Following the procedure suggested by Hubley-Cozey et al. (2006) a moving window algorithm was used to determine the 0.1 second window during which the maximum EMG amplitude occurred. This maximum value (averaged across the 0.1 second window) was then take as the highest MVIC value for that muscle. This procedure was identical for each muscle and each subject and the maximal values exported to excel for subsequent analysis. Using these data, I then identified, for each muscle, the limb position corresponding to the largest EMG signal, i.e. the position which would be most appropriate to use for an MVIC for that muscle/individual. This analysis was repeated for all muscles/participants in order to obtain data identifying the most appropriate MVIC across the different muscles.

Results:

This data showed that, for the biceps femoris, the position most commonly associated with a maximum contraction was the position of 55° knee flexion with external rotations (Table 3-4). This position was optimal for 42% of healthy individuals and 50% of patients with knee OA exercise. Similarly, for semitendinosus, 67% of healthy subjects and 44 % of individuals with knee OA (Table 3-4) exhibited the highest MVIC at this position (55° knee flexion with external rotation). For the quadriceps muscles, the highest MVIC for both the medial and lateral quadriceps was in 15 degrees of knee flexion with slight internal rotations (Table 3-4). Specifically, the percent of individuals who achieved the highest MVIC for the VM in this position was 44% for the knee OA group and 42% for the healthy group, while for the VM muscle, it was 44% and 33 % for individuals with knee OA and healthy participants respectively (Table 3-4). For the gastrocnemius muscle, the data showed that over 80% of participants generated a maximal contraction in the standing position (PF stand) (Table 3-4).

Table 3-4. The maximal voluntary isometric contraction exercises MVIC Standard positions and the numbers of subjects for both healthy and knee OA groups.

		Healthy n=12		OA n=16	
		Number of subjects	%	Number of subjects	%
Medial Gastrocnemius	PF Sit	1	8.33	0	0.00
	PF Stand	11	91.67	16	100.00
Lateral Gastrocnemius	PF Sit	2	16.67	2	12.50
	PF Stand	10	83.33	14	87.50
Vastus medialis	KE 15 ER	2	16.67	4	25.00
	KE 15 IR	5	41.67	7	43.75
	KE 45 ER	2	16.67	5	31.25
	KE 45 IR	3	25.00	0	0.00
Vastus lateralis	KE 15 ER	4	33.33	2	12.50
	KE 15 IR	4	33.33	7	43.75
	KE 45 ER	1	8.33	3	18.75
	KE 45 IR	3	25.00	4	25.00
Semitendinosus	KF 15 ER	1	8.33	0	0.00
	KF 15 IR	2	16.67	3	18.75
	KF 55 ER	2	16.67	6	37.50
	KF 55 IR	8	66.67	7	43.75
Biceps femoris.	KF 15 ER	3	25.00	3	18.75
	KF 15 IR	2	16.67	2	12.50
	KF 55 ER	5	41.67	8	50.00
	KF 55 IR	1	8.33	3	18.75

Summary,

Given the results described above, I adopted the following positions for MVIC testing:

- For both of the quadriceps muscles (VM and VL): KE15 IR (knee flexed to 15°and slightly internally rotated)
- For the biceps femoris: KF55 ER (knee flexed to 55°and slightly externally rotated)
- For the semitendinosus: KF55 IR (knee flexed to 55°and slightly internally rotated)

- For the medial and lateral gastrocnemius: PF stand (plantarflexion of the ankle against body weight in standing).

3.6.6 Further justification for the use the MIVC normalization method

As explained in section 3.6.4, it is necessary to select an appropriate method to normalise the EMG signals. This is because EMG magnitude is dependent on factors other than the level of muscle activation, such as skin conductivity, thickness of subcutaneous fat and number of motor units in the EMG collection volume (De Luca, 1997). In section 3.6.4, a theoretical justification was provided for the use of MVIC normalisation. However, to provide further support for the use of MVIC data, a comparison of MIVIC data was performed between the healthy group and the group with knee osteoarthritis. For this analysis, MVIC contraction data for each muscle, in the test position specified above, were compared between the two groups The MVIC value, for each muscle, was obtained using the linear envelop approach described in the previous subsection (Section 3.6.5) and is presented below in the units of MicroVolts. For each muscle, a comparison between the two groups was performed using an independent t-test.

Table 3-5. Differences in maximal voluntary isometric contraction exercises (MVIC) for selected positions of the muscle around the knee joint. Mean(SD) data is expressed in MicroVolts.

Muscle	Group	Mean (SD)	P- value
MG (PF stand)	**Healthy**	**212.37 (97.10)**	**0.16**
	OA	**176.48 (74.07)**	
LG (PF stand)	**Healthy**	**198.09 (92.21)**	**0.76**
	OA	**191.11 (60.84)**	
VM (KE15 IR)	**Healthy**	**137.18 (74.40)**	**0.28**
	OA	**115.19 (63.62)**	
VL (KE15 IR)	**Healthy**	**135.60 (60.13)**	**0.29**
	OA	**114.38 (72.81)**	
ST (KF55 IR)	**Healthy**	**236.43 (112.14)**	**0.64**
	OA	**222.39 (93.42)**	
BF (KF55 ER)	**Healthy**	**203.29 (92.62)**	**0.86**
	OA	**199.08 (76.85)**	

The results, shown in Table 3.5, illustrate that while differences exist between values for the two groups, none of these differences were significant. Percentage difference between the healthy and knee OA subjects were approximately 20% higher in the healthy group for the medial gastrocnemius but less than 20% for the other muscles. Importantly, the comparison of muscle activation levels between participants with knee OA and healthy controls (described in detail in Section 3.6.5) showed that five out of the six muscle studied were significantly higher in the group with knee OA. Although it is possible that this difference could have been due to higher MVIC levels in the healthy groups, the differences observed in section 4.4.4 are typically 50% larger in the group with knee OA. Therefore, given the differences of less than 20% in the table above, it is unlikely that differences in MVIC level could explain the observed differences in muscle activation. These difference most likely reflect true differences in muscle coordination between the two groups.

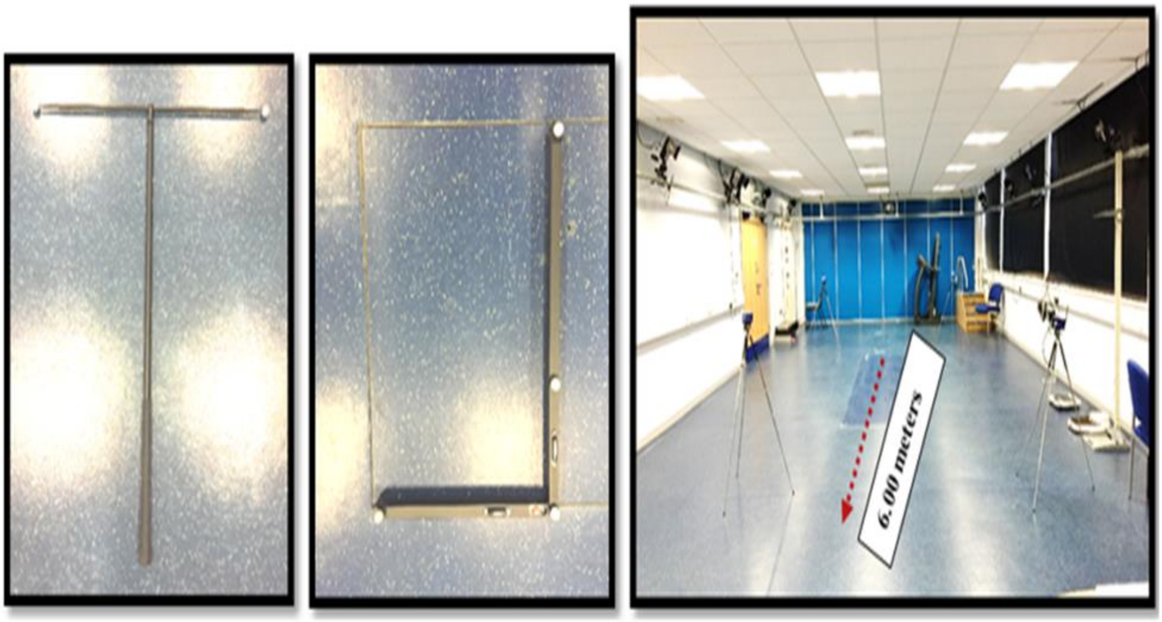
The approach of normalisation of EMG to MVIC was selected based on a consideration of its previous use, strengths and weaknesses. First, it is noted that this approach is well established for studies of knee OA, as evidenced by Mills, Hunt, Leigh and Ferber's (2013) systematic review and meta-analysis of fourteen studies to identify frequently-occurring neuromuscular changes with knee OA and the impact of the level of OA, joint laxity and varus alignment. The majority of the studies included in the review followed this approach. Good reliability is reported for MVIC (Halaki & Ginn, 2012; Rutherford, Hubley-Kozey & Stanish, 2010; Lee & Jo, 2016). However, Halaki and Ginn (2012) emphasize the need to achieve maximal neuromotor activation through using the most appropriate MVIC test for each muscle to be measured. The findings above of only small, non-significant, differences between the MVIC values, provides confidence that the maximal neuromotor activation was achieved with the protocol used for this thesis.

3.7 Kinematic and kinetic data collection procedures:

3.7.1 Calibration and set up of the motion capture system

Calibration of the camera system was required for collection of kinetic and kinematic information. For this, a reference item (A frame of metal in an L-shape) was located on force platform first on a corner, aligned parallel with both Y and X axes and marked with 4 markers which are positioned a set length from the force platform co-ordinate origin (corner) figure 3-8 b). These were calculated on an automatic basis and entered into the computer program (Winter, 2009). The medial/lateral (X) axis, anterior posterior (Y) axis and vertical (Z) axis act together with the reference item to form the co-ordinate system origin for the lab. In order to calibrate the system, random motions are made for one minute's duration using a wand to which 2 markers are attached Figure 3-8 A, with the reference object remaining situated on the force platform. This process establishes where the sixteen cameras are positioned and oriented in relation to the system of co-ordinates established in the laboratory (Payton and Bartlett, 2008). Following this calibration process, the errors in identifying the 3D coordinates during the calibration procedure were 1 mm.

With 3D marker capture and kinematic analysis, at least 2 cameras must be able to image an individual marker for the software to be able to reconstruct the precise 3D co-ordinates (Cappozzo et al., 2005). During the piloting phase of the experimental work, I checked each marker in the chosen configurations to be sure that it could always be seen (throughout the collection volume) and therefore the 3D co-ordinates calculated. With 3D motion analysis, the body is segmented into a number of rigid segments, each of which is tracked separately. In order to define the precise position and orientation of each segment, at least three non-colinear markers must be attached to each segment. Provided all three markers can be tracked accurately, then specialised software can be used to calculate 3D position and orientation of each segment both relative to the reference, (laboratory) system and also relative to other segments. By combining the kinematic information with precise information on forces under the foot, inverse dynamic calculations can be used to derive moments at the ankle, knee and hip (Winter et al., 1990). Both the kinematic and the kinetic calculations were performed using the Visual 3D software (see later section).



A) B) C)

Figure 3-8. A) Calibration wand with two markers. B) L-shape of Calibration frame with four markers. C) The sixteen-camera Qualisys Oqus motion analysis system, force platforms walking distance and direction at gait lab.

3.7.2 Marker placements

A set of 14.5 mm-diameter reflective markers were adhered to the skin by means of hypoallergenic tape stuck on the flat base of the markers. As explained above, a minimum of 3 markers, which are not aligned linearly, are needed per rigid body segment to describe the 3D position and orientation (Cappozzo et al., 1996). According to Cappozzo et al. (1996) artefacts caused by motion of the skin moving markers in relation to the bones presents a significant source of errors in gait trials. For this reason, the CAST (Calibrated Anatomical

System Technique) was applied to reduce the effect of these artefacts, which utilises rigid cluster plates (Cappozzo et al., 1996, Cappozzo et al., 1995). In CAST, the POSE (position and orientation of body segments) is derived by placing individual anatomical markers at points of bony prominence to define rigid segments, while clusters of technical markers are used to track segment motions. By using this approach, artefacts from skin movement were minimized (Cappozzo et al., 1996).

Cappozzo et al. (1997) identified the optimal marker numbers to be included in the cluster for a segment in order to give the most accurate orientation and positioning data within practical parameters. They suggested clusters of 4 markers attached in locations to minimise global skin motion artefacts, using anti-migratory self-attaching tape, and that tracking clusters should not be placed over bony prominences, to reduce displacement effects. Manal et al. (2000) reported the use of a rigid plate for marker clusters rather than mounting them directly on the skin, therefore reducing skin movement artefact occurrence as no motion is possible within a cluster. Given these recommendations, I used 4-marker clusters mounted on rigid plates made of polypropylene, and attached these to the two shanks, two thighs and pelvis with double-sided adhesive tape, before wrapping the clusters in Fabriofoam Superwrap™.

CAST was selected over the set of markers put forward by Hayes, because the latter set create the potential for error to be transferred along the lower limbs, with incorrect location of markers for the pelvis impacting other marker placement. Further, the Hayes approach is more open to soft tissue artefact production, with both wands and markers mounted directly onto skin. I adopted a CAST, seeking to minimise the presence of artefacts from motion of the skin by placement of markers centrally within a body segment. Further, the way in which markers on joints are oriented and positioned should be constant, and mathematical calculation of the relationship of the markers on segments to markers on joints should be achievable, in static calibration. After removal of the calibration markers from the joints the segment, this allows for the use of tracking marker data to calculate segmental position and orientation.

3.7.3 Kinematic and kinetic data collection

Kinematic/kinetic data was collected using a Qualisys motion analysis system and two AMTI force platforms. Sixteen infrared cameras (Qualisys, Sweden) were used to capture the 3-dimensional positions of retro-reflective markers (see Figure 3-9) attached at various anatomical positions on the lower limbs and trunk. The small reflective markers were stuck to the skin using hypo-allergenic adhesive tape. Individual markers were attached on anterior superior iliac spines (ASISs), posterior superior iliac spines (PSISs), iliac crests, right greater trochanter, left greater trochanter, lateral femoral epicondyles, medial femoral epicondyles, lateral malleoli, medial malleoli, the 1st, 2nd and 5th metatarsal heads and calcaneal tubercle. In addition, a cluster pad (as explained above) with four markers was placed on the shank, thigh and pelvis using bandages. Markers were also placed on the anatomical joint landmark of trunk, Jugular notch, Thoracic vertebrae T2 and T8, shoulder (Right and left side acromion process). The rationale for this choice of marker placements is discussed in detail in section 3.10.2 below.

3.8 Footwear

Once all EMG electrodes (Section 3.6.3) and reflective markers (Section 3.7.2) were in place, then the gait analysis data was collected in two different types of footwear:

1. Control footwear which was a standard oxford style shoe (figure 3-9)
2. Rocker footwear which was the three-curved rocker shoe, described below.

Figure 3-9. Control shoe (left pair shoes for women and right pair for men).

The control shoe used was a high street Oxford shoe, which this was selected as a standard shoe as it is widely available and is in widespread use. The outsole has a low heel, and the shoes are lace-up. While this type of shoe is less commonly worn by women, the shoe design for men and women was kept constant for all participants to avoid creating a factor for variation in the results. See Figure 3-10.

3.8.1 The three-curved rocker shoe

Rocker shoes are designed with a rigid outsole, which is designed to "rock" the foot forwards during the stance phase of walking. In a recent study, Hutchins et al. (2012) proposed a rocker shoe design with a rocker profile created from three different curves, generated from circles drawn around the hip, knee and ankle. The motivation for this design is to redirect the GRF vector so that it is closer to the respective joint centres and therefore reduces the effective distance between the GRF vector and the joint centre, thereby reducing joint moments (Figure 3-13 below). Although further research is required to fully validate the proposed effects of the three-curved rocker shoe, Hutchins et al. (2012) suggests that the realignment of the GRF vector will reduce joint moments and therefore reduce muscle activation around the hip, knee and ankle.

According to Hutchins et al. (2012) this design of shoe will modify the direction and orientation of the GRF during early stance and this will bring about a corresponding reduction in gastrocnemius activation. In-turn this will produce a corresponding decrease in gastrocnemius-quadriceps muscle co-contraction during early stance and therefore may reduce co-contraction in people with knee OA. Hutchins et al. (2012) also suggest that the three-curve rocker shoe will reduce the hip moment, as a result of the change in the direction of the ground reaction force vector, during the early-mid stance part of the gait cycle. Furthermore, as explained above, it has been suggested that the destabilising effect of the rocker shoe may improve postural alignment by encouraging a better alignment of body segments during gait. These effects, of reducing forward trunk inclination, combined with a reduction in hip moment, due to the redirection of the ground reaction force, are likely to lead to decreases in hamstring muscle activity and therefore reductions in hamstring-quadriceps co-contraction. Given the mechanisms (explained above) by which this three-curved rocker shoe could reduce both hamstring-quadriceps and also gastrocnemius-quadriceps co-contraction, it was deemed the most appropriate design for this study.

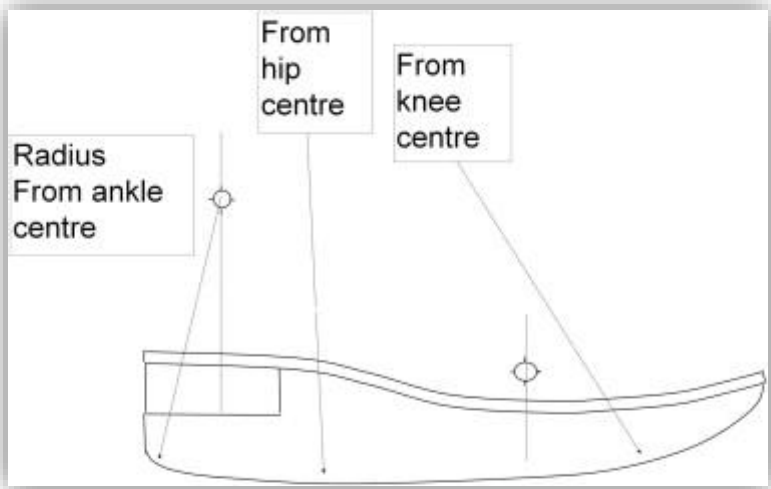


Figure 3-10. The three radiuses of 3 curve rocker shoe (T3C)

Figure 3-11. The three-curved rocker shoe (T3C) (left pair for women and right for men).

3.9 Gait analysis procedures

For each shoe, a static trial was collected initially followed by at least 5 successful dynamic walking trials. For the static trial, the subject stood stand on the force plate form for 10 seconds while motion capture (kinematic) data was collected. Following the static trial, each participant was given a 5-minute familiarisation period with each footwear type and then was asked to make at least five successful walking trials. Successful trials were those in which subjects made contact with the force plate with the most affected leg for knee OA subjects, and the matched leg for the healthy volunteer, without targeting and with appropriate and consistent gait pattern. For the gait trials, walking was undertaken at a self-selected speed for knee OA patients. However, people with knee OA typically walk slower than health controls (Kaufman et al., 2001). Therefore, in order to ensure that the results were not confounded by speed, healthy subjects were instructed to walk slowly. During each trial, walking speed was measured using optical timing gates and subsequent comparison of the speeds between the two groups showed that the knee OA group walked with a mean (SD) speed of 1.1 (0.13) m/s and the healthy group with a mean (SD) speed of 1.09 (0.13) m/s. An independent t-test was carried out to analyse the speed difference and showed that there was no significant difference between the groups ($p=.81$). To ensure that subjects maintained the same walking speed in each of the footwear conditions, walking speed was measured using optimal timing gates after each trial and only trials within +/-10% were accepted.

Following successful completion of the walking trials, the subject then repeated the process for the next pair of shoes (order randomised), allowing sufficient time to get used to each of the different footwear conditions. During each of the gait trials, kinematic data and kinetic data were collected with the Qualisys motion analysis system (100Hz) and two AMTI force platforms (1000 Hz) respectively, while a wireless DTS -EMG system (1500 Hz) was used at the same time for recorded muscle activity, as explained in Sections 6 and 7 respectively. Following the gait data collection, the MVIC data was obtained as explained in Section 3.6.5 above.

3.10 kinemetic and kinetic data processing

3.10.1 Choice of biomechanical model

A widely-used rigid body model is the CGM (conventional gait model) (Davis, Ounpuu, Tyburski, & Gage, 1991; Kadaba, Ramakrishnan, & Wootten, 1990). However, with this approach, it has been suggested that the use of minimal, but widely spaced, markers can increase the sensitivity and effectiveness of CGM as the skin moves (Cereatti, Camomilla, Vannozzi, & Cappozzo, 2007). Furthermore, the CGM does not allow identification of the way in which a segment is positioned and orientated without reference to another segment, as there are only two markers tracking each segment (Cereatti et al., 2007; Schwartz, Trost, & Wervey, 2004). As an alternative to the CGM, Della Croce (1996) propose a ‘6 degrees of freedom’ (6DOF) model which allows segments to be individually tracked by attaching 4 retro-reflective markers to each segment. In this model, transverse, frontal and sagittal rotation variables are derived from the data alongside anterior-posterior, medial-lateral and vertical translational parameters for every joint, forming a picture of how the joint is oriented and located in three dimensions. This allows independent measurement of all segments, minimizing error rates. The 6DOF approach is reported to reduce error in comparison with CGM (Cappozzo et al., 1996; Cereatti et al., 2007). Given the potential advantages of the 6DOF approach, it was chosen in this thesis as the basis for calculating joint kinematics and moments.

3.10.2 Definition and tracking for the thoracic segment

The trunk weighs more than any other body segment (Gillet et al., 2003), and therefore the kinematics of the trunk need to be considered carefully when analyzing gait (Romkes et al., 2007). However, a range of different approaches have been used for modelling the thoracic segment, each with a different set of markers. For example, Davis et al. (1991) used just two markers to define this segment, positioned between the left and right clavicle and at C7. In another study, Nguyen and Baker (2004) and Gutierrez, Bartonek, Haglund-Åkerlind, and Saraste (2003) also use these two markers, but add two additional markers at T10 and on the sternum (Gutierrez et al., 2003). Another approach proposed by the International Society of Biomechanics (ISB) uses markers placed at the IJ (incisura jugular), XP (process of xyphoid) C7 and T8.

In a recent study Armand et al. (2014) sought to determine the optimal marker set for tracking thorax motions during clinical gait analysis, comparing different marker sets. The authors first studied a set of markers incorporating incisura jugularis (IJ) and the processus xyphoid (XP), with a low thorax marker at T6, T8 or T10. This set was compared to a set consisting of IJ, and T2 or C7 along with T8/T10. The former set appeared to lead to excessive amplitude of movement in the frontal plane and therefore the latter set was deemed more appropriate. From this latter set, the C7 marker was rejected for analysing walking patterns as it was found to be highly responsive to movements of the head. Both T8 and T10 were found to give good reliability as markers. Overall, the authors found suggested that the optimum marker set for tracking the thoracic segment was IJ, T2 and T8 or T10 (Armand et al., 2014), and these recommendations are used in the thesis.

3.10.3 Defining and tracking the different segments in the model

The biomechanical model used

A three-dimensional 'six degrees of freedom' (6DOF) model was constructed as biomechanical model, comprising 8 rigid segments: the thorax, pelvis, right and left thigh, right and left shank and right and left foot (Figure 3-12). Each segment was defined and tracked separately following the 6DOF modelling approach. As explained above, each segment was tracked with at least 3 (and sometimes 4) non-colinear markers which enabled the position and orientation of each segment to be precisely determined relative to the laboratory origin. With these data, biomechanical software (see Section 3.10.5) was used to derived joint kinematics. The biomechanical model is described in Table 3-6 below.

Table 3-6. Biomechanical model segments and tracking markers

Segment	Proximal joint	Distal joint	Tracking markers
Trunk	Right and left grater trochanter	Right and left acromion	Jugular notch(IJ), 2nd 8th Thorax vertebra Thorax vertebra
Coda Pelvis	Right and left - anterior superior iliac spine	Right and left - posterior superior iliac spine	Pelvis cluster belt (3 markers)
Thigh	Right and left greater trochanter	Medial and Lateral femoral epicondyles	Thigh cluster (two clusters; 4 markers for each)
Shank	Medial and lateral femoral epicondyles	Medial and lateral malleolus	Shank cluster (two clusters; 4 Markers for each)
Foot	Medial and lateral malleolus	first and fifth foot metatarsals	Right and left foot 1st 2nd 5th metatarsal head metatarsal head Right and left heel calcaneus

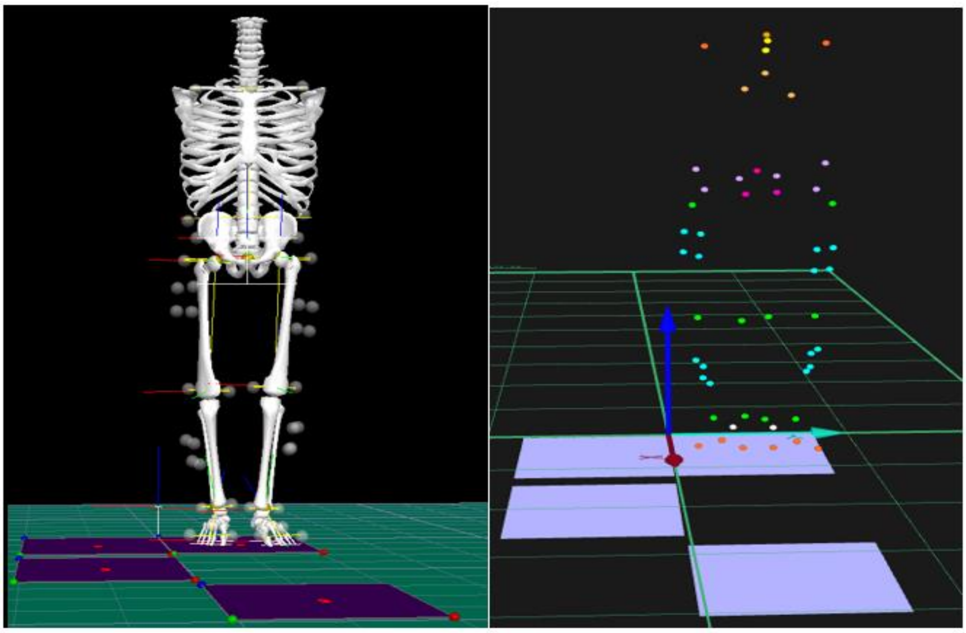


Figure 3-12. (A) Visual 3D biomechanical Model, (B) Qualysis Track Manager

3.10.4 Raw data pre-processing

The Qualisys motion analysis system (Qualisys, Gothenburg, Sweden) and two AMTI force platforms (AMTI BP400X600, AMTI, USA) were used to acquire kinematic and kinetic data respectively. Kinematic data acquisition was made at 100 Hz, while kinetic data was obtained using two force platforms at 1000 Hz. Once collected, the raw data for kinematics were interpolated to correct for tracking errors (max 10 frames) low pass filtering used to remove the high frequency noise from the marker trajectory data. The filtering was performed, in Visual 3D, using a Butterworth 4th order bi-directional filter with a cut-off point of 6Hz for kinematics (Winter, 2009). The force data was also low pass filtered at 25Hz to remove any high frequency noise (Schneider and Chao, 1983), again in Visual 3D.

3.10.5 Kinematics and kinetic calculations

Following filtering, the raw data was used as input to the 6DOF biomechanical model (described above) in order to derive segmental positions and orientations. Visual 3D was then used to implement Euler angle calculations to determine 3D joint angles between each set of adjacent segments. Force data was the used as part of inverse dynamic calculations, to derive joint moments at the hip, knee and ankle and to derive CoP in a foot coordinate system. All moment data was subsequently normalised to body weight.

Specific kinematic and kinetic curves were then selected for subsequent analysis, including hip, knee and ankle sagittal plane kinematic and moment data, thorax (relative to the laboratory) kinematic data and CoP data. All kinematics, kinetic and CoP data curves were time-normalised (0-100%) to the stance phases of the gait cycle, from initial contact (heel strike) to toe-off using the force data and a threshold of 5N. Finally, all gait curves were exported from V3D to Microsoft Excel 2016. This was done for each individual trial and each shoe, creating a database containing all data. Using these data, the ensemble averages from the kinematic, kinetic and CoP data were then calculated across stance phase for each subject and each shoe. From these curves, specific outcomes were calculated for each of the separate studies. Further details are provided in Chapters 4, 5 and 6.

3.11 EMG data processing

Following data collection, the gait EMG data was exported in a c3d format which could be ready by a custom Matlab programme which was used to perform standardised EMG processing. This processing involved using 20Hz high pass FFT filter to remove noise and movement artefact. The signal was then rectified and a low pass filter (6Hz Butterworth) used to create a linear envelop. Note this cut frequency has been used previously in earlier EMG studies of people with knee OA (Winter 1990; Hubley-Kozey et al. 2006). The EMG signals were then time normalised to the stance phase of the most affected limb using gait event data captured from the force platforms. Following EMG processing, the data were exported to a Microsoft Excel 2016 so that ensemble average curves could be produce for each muscle, shoe and participant.

The MVIC data processing, described in the MVIC pilot study (Section 3.6.5), was used to obtain a reference value for each muscle for each person. This MVIC value was then used to normalise the EMG data for each muscle individually from the ensemble average of the walking trials. Specific outcomes, relating to each research question, were then derived from these ensemble average normalised curves, either for individual muscles or by combining agonist-antagonist pairs characterise co-contraction. Previous researchers have suggested two possible methods for calculating co-contraction from pairs of muscle groups. The first of these methods, suggested by Heiden et al. (2009), is based on a co-activation ratio, calculated from the angonist and antagonist muscles, and is defined as follows:

$$\text{if} \quad \text{mean}(\text{Antagonist}_i)) < \text{mean}(\text{Agonist}_i) : \cdot \text{CCR}_i = 1 - \frac{\text{mean}(\text{Antagonist}_i)}{\text{mean}(\text{Agonist}_i)}$$

$$\text{else} : \cdot \text{CCR}_i = \frac{\text{mean}(\text{Agonist}_i)}{\text{mean}(\text{Antagonist}_i)} - 1$$

The other method is to simply sum the activity of the agonist and antagonist at each point of the gait cycle to produce an overall ensemble average co-contraction curve (Winby et al., 2013, Preece et al., 2016). Importantly, a recent modelling study showed that this latter approach of summing the activity correlated more strongly with knee loads (Winby et al., 2013) and therefore was used throughout this thesis. In chapters 4,5 and 6, muscle activity or co-contraction were averaged across a specific window of the gait cycle to create appropriate outcomes. Further details of these outcomes are provided in Chapters 4, 5 and 6.

3.12 Reliability study:

As this thesis aimed to investigate gait kinetics, kinematics and EMG, it was necessary to ensure that the data produced was reliable, and repeatable between different sessions. Therefore, before the main data collection began, a reliability study was carried out.

Aim: To assess the test-retest repeatability of the gait kinetics and kinematics and muscle activity between days.

Method

The sessions took place at the Gait Laboratory of the University of Salford. Three healthy subjects were recruited for this study. The participants were postgraduate students at the University of Salford. The full lab protocol, explained above , was performed on two occasions separated by at least 2 days. However, due to a problem with a development of the EMG testing protocol, MVIC data were not collected. EMG data was therefore normalised to the peak value (the peak dynamic method) rather than the MVIC.

The Coefficients of Multiple Correlation (CMC), which measures the similarity of waveforms, was used to quantify the between day measurements for each curve (Kadaba et al., 1989). The CMC value can be any number from zero (0) to positive one (+1). The higher the reliability (waveform match), the closer the result is to one. According to Growney et al. (1997), similar waveforms with values of more than 0.8 demonstrate high test-retest reliability. All statistics were performed using The Statistical Package for the Social Sciences (SPSS 20, IBM, New York, USA).

Results:

Three male healthy subjects in the study; mean age 37.33 (1.5) years; age range 36-39 years with mean 37.3 (1.5); mean height 1.653 (.01)m; height range 1.64-1.67 cm; mean mass 74.3(12.5) Kg; weight range 60- 83 Kg) mean 74.3(12.5).

Test-retest reparability of EMG during gait.

The results demonstrate high levels of repeatability in kinematic, kinetic and EMG measures between sessions on different days. CMC results for trunk inclination (Figure 3-13 to 3-15), moments, EMG (medial and lateral gastrocnemius, VM, VL, semitendinosus and biceps femoris activity) for all subjects are presented in Table 3-7.

Table 3-7. Within- & between-days CMC

Variable		Subject	CMC
Kinematic	**Trunk inclination (°)**	1	0.97
		2	0.97
		3	0.94
Kinetic	Hip moment	1	0.99
		2	0.99
		3	0.99
	Knee moment	1	0.99
		2	0.99
		3	0.99
	Ankle moment	1	0.99
		2	0.99
		3	0.99
EMG	Medial gastrocnemius activity	1	1.00
		2	0.99
		3	1.00
	Lateral gastrocnemius activity	1	0.99
		2	0.99
		3	0.99
	VM activity	1	1.00
		2	1.00
		3	1.00
	VL activity	1	0.99
		2	0.99
		3	0.99
	Semitendinosus activity	1	0.99
		2	0.99
		3	0.99
	Biceps femoris activity	1	0.99
		2	1.00
		3	0.35

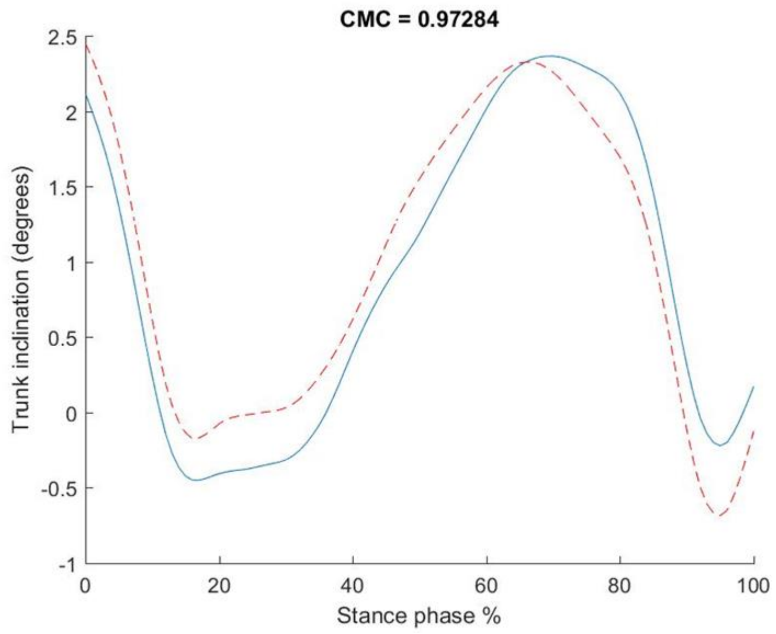


Figure 3-13. Subject 1: trunk inclination with the dotted line is the second day and the solid line is the first day

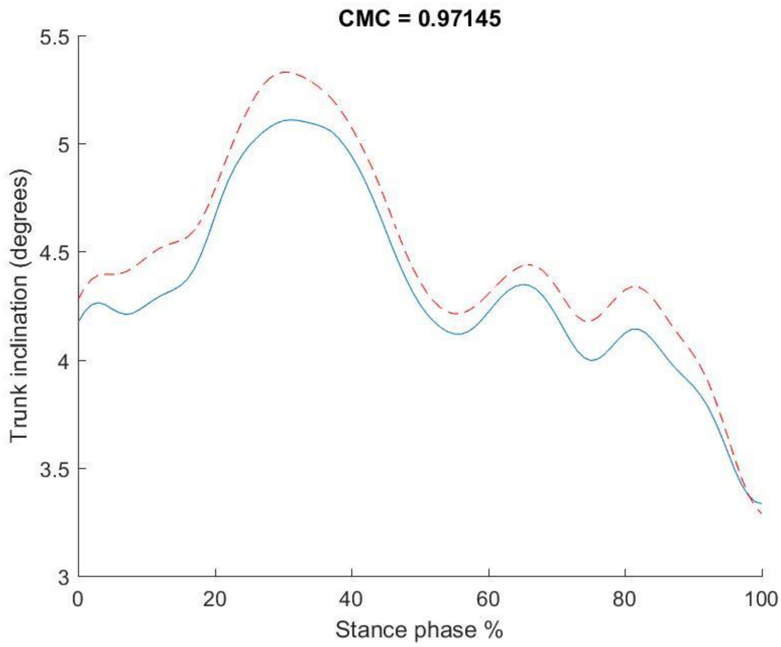


Figure 3-14. Subject 2: Trunk inclination with the dotted line is the second day and the solid line is the first day.

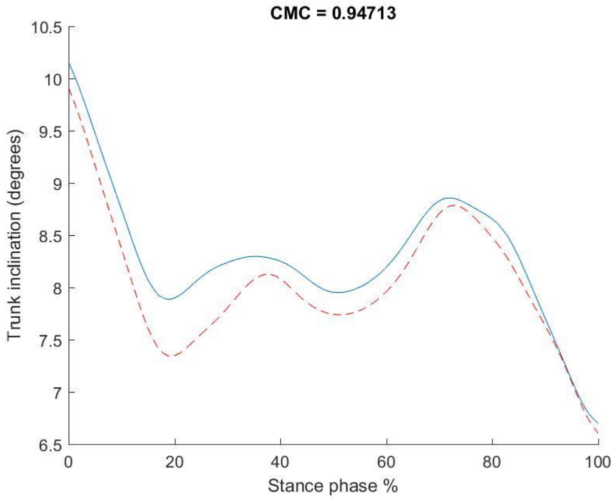


Figure 3-15. Subject 3: Trunk inclination with the dotted line is the second day and the solid line is the first day.

As can be seen above, the study showed very good repeatability in all outcomes. Specifically, CMC values of .97, .97 and .94 were observed for the three subjects for trunk inclination, while for the remaining variables, values of 0.99 - 1 were obtained. Sample waveforms shown in Figures 3-13, 3-14 and 3-15 for trunk inclination, show that values for this variable closely matched at each point during the stance phase. These data provide confidence in my ability to perform the laboratory protocol repeatedly.

Chapter 4 - Study One: Trunk inclination in people with knee OA

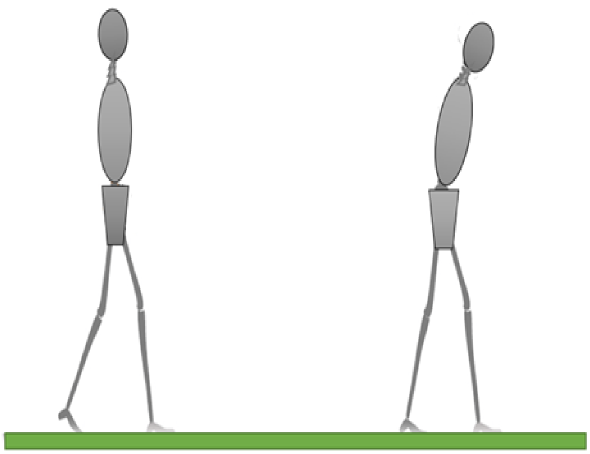
Overarching aim:

To fully characterise differences in sagittal plane trunk inclination and other biomechanical variables between people with knee OA and healthy controls.

4.1 Overview of the study

Previous research has shown that people with knee OA stand with increased trunk inclination (Figure 4-1) (Turcot et al., 2015b). However, there is no definitive data on trunk inclination in people with knee OA, and it is not clear whether the trunk lean adopted in standing is maintained in walking. Therefore, the first three research questions (RQ1A-1C) in this chapter explore trunk inclination during walking and standing in people with knee OA.

Figure 4-1. Walking with normal trunk position on the left and with forward trunk inclination on the right.



If patients with knee OA do walk with an increased inclination of the trunk, then this may lead to a change in the CoP path under the foot. However, there appear to be conflicting results in previous research, with some studies showing a definite alteration in the CoP path (Saito et al., 2014) in people with knee OA and others showing no differences in ankle plantarflexor moments (Astephen et al., 2008). These conflicting findings of previous research motivated research questions RQ1D-1E, which seek to explore whether alterations in CoP path are a characteristic of OA gait and whether such alterations are linked to variations in forward trunk lean.

Previous research has consistently shown that people with knee OA walk with increased muscle activations (Zeni et al., 2010; Childs et al., 2004; Hortobagyi et al., 2005; Hodges et al., 2016) and an increase in the associated co-contraction (Childs et al., 2004, Lewek et al., 2004; Hodges et al., 2016; Hubley-Kozey et al., 2006). Furthermore, research has shown that knee OA to be associated with increased hip extensor moment (Liu et al., 2014) and, in some cases, alterations in knee moments *(*Kaufman et al., 2001, Astephen et al., 2008, Baliunas et al., 2002a, Sritharan et al., 2016; Mündermann et al., 2005). Therefore in the final set of research questions in this chapter (RQ1F-1H, I have repeated previous research which has sought to characterise the differences in moments, muscle activations and co-contraction between people with knee OA and healthy control subjects. The purpose of this investigation was to fully characterise the cohort and to facilitate comparisons with previous research.

4.2 Research Questions

The research questions for this study are as follows:

1. RQ 1A: Do individuals with knee OA walk with an increased inclination of the trunk?
2. RQ 1B: Do individuals with knee OA stand with an increased inclination of the trunk?

3. RQ 1C: Does trunk inclination in standing correlate with trunk inclination in walking both in a group of individuals with knee OA and also in a healthy cohort?
4. RQ 1D: Is there a difference in CoP between healthy and knee OA subjects?
5. RQ 1E: Is there a link between forward trunk inclination and anterior shift of CoP?
6. RQ 1F: What are the differences in hip/knee/ankle moments between healthy and knee OA subjects?
7. RQ 1G: What are the differences in hamstring/quadriceps/gastrocnemius muscle activity between healthy and knee OA subjects?
8. RQ 1H: What are the differences in co-contraction of between healthy and knee OA subjects?

For each of these studies, I chose to focus on the period 15-25% of the stance phase as this has been shown to the period of peak loading during walking. This idea was discussed in detail in Section 2.4.5 of the literature review section.

4.3 Methodology

4.3.1 Sample and population

Data for all the three studies were collected during a single visit from two groups; healthy (n=20) and knee OA (n=27) subjects across both genders. Further details of the specific inclusion/exclusion criteria are provided in Section 3.2 of the methods section.

4.3.2 Derivation of outcome measures

In the methods chapter, a full description was provided for all the experimental testing and initial data processing required to address the eight research questions stated above. The methods section explains how specific joint kinematics, moments and CoP were derived and also details the post processing which was applied to the raw EMG data. In the section below, I have presented a detailed explanation of how the specific outcomes (relating to each research question) were derived from the kinematic, moments, CoP and EMG data. As explained in Chapter 3, only data collected whilst participants wore the control shoe was used to address the questions in this chapter.

RQ 1A: Do individuals with knee OA walk with an increased inclination of the trunk?

Trunk inclination was calculated as the orientation of the thoracic segment (see methods section) with respect to the laboratory reference frame in the sagittal plane. For each person, an ensemble average trunk inclination was calculated over the stance phase of gait and from this ensemble average, a value for mean trunk lean was calculated over the specific period of stance phase (15-25%). Mean trunk lean (over 15-25% stance) was the outcome used to address this research question.

RQ 1B: Do individuals with knee OA stand with an increased inclination of the trunk?

As explained in the methods section, each participant was required to stand still for a period of at least 10 seconds whist kinematic data was collected for the calibration trial. A one second window (selected on the basis of there being minimal body movement) was used to obtain a representative signal for trunk inclination during standing. Mean standing trunk inclination was then calculated across this one-second window and used as the outcome to address this research question.

RQ 1C: Does trunk inclination in standing correlate with trunk inclination in walking both in a group of individuals with knee OA and also in a healthy cohort?

The outcomes defined above (RQ 1A and 1B) were used to address this question.

RQ 1D: Is there a difference in CoP between healthy and knee OA subjects?

Ensemble average CoP signals were derived, in the anterior posterior (A-P) direction, for each individual and then a mean CoP position calculated over the period 15-25% stance. This mean CoP was the outcome used to address this research question.

RQ 1E: Is there a link between forward trunk inclination and anterior shift of CoP in the specific period of gait?

The outcomes were explained in RQ 1D

RQ 1F: What are the differences in hip/knee/ankle moments between healthy and knee OA subjects?

Ensemble average curves for sagittal hip, knee and ankle moment were calculated for each participant. Then, to produce a single outcome for each curve, we calculated the mean over the specific period of stance phase (15-25%) during walking.

RQ 1G: What are the differences in hamstring/quadriceps/gastrocnemius muscle activity between healthy and knee OA subjects?

The means of biceps femoris, semitendinosus, vastus medialis, vastus lateralis and medial and lateral gastrocnemius activity, over 15-25% stance were calculated to answer this question. For each individual, we created a smoothed linear envelop signal which was normalised by the MVIC signal (as explained in the methods section) for each muscle over each walking trial.

Mean ensemble EMG profiles were calculated across the time window 15-25% stance, giving a value of muscle activation over the period of interest for each muscle in every individual.

RQ 1H: What are the differences in muscle co- contraction between healthy and knee OA subjects?

In order to quantify co-contraction, I used a method that involved summing the activation levels of the agonist and antagonist muscles. This was done separately for the medial muscles and the lateral muscles. Specifically, two quadriceps–hamstrings co-contraction outcomes were obtained along with two quadriceps–gastrocnemius outcomes (one for the lateral muscle and one for the medial muscle pairs) over the time window 15-25% stance (Heiden et al., 2009b), as this corresponds to peak loading, as explained in details in section 2.4.5.

4.3.3 Statistical analysis

All statistical analysis relating to research questions RQ 1-H were performed using the statistical package for social studies (SPSS) version 23 for Windows. This study involved one independent variable, the tested group; and evaluated sixteen tested dependent variables. Before the results were analysed, data screening was carried out for normality and homogeneity of variance assumptions, to allow parametric calculations of analysis of difference. No outliers were found when a box and whiskers boxplot was carried out for each tested variable. Descriptive analysis using histograms with normal distribution curves revealed normal distribution of data each variable, and normality tests did not suggest significant differences between the distribution of the tested sample's raw data and normally-distributed population data, using the same means as those of the tested variables ($p>0.05$) and evaluated using the Shapiro-Wilk's test. Homogeneity of variances was identified using Levene's test for equality of variances testing ($p>0.05$) for each dependent variable. Thus, the results did not violate the parametric assumption, and parametric data analysis was possible. Data are given in the form of mean ± standard deviation (SD) in the descriptive statistics (see following section).

In order to investigate the differences between knee OA and healthy groups (research questions RQ 1A, B, D, F, G, and H) specific outcomes, defined above were used. These were forward trunk inclination, moments, A-P CoP, each muscle's EMG profile and co- contraction averaged across the time window 15-25 % stance. These comparisons were calculated using an independent t-tests to determine if there were any significant differences in the mean values of each the outcome between the groups.

A Pearson's correlation coefficient (r) was used to investigate the relationship between forward trunk inclination and the outcomes defined for research questions RQ 1C and RQ 1E (standing trunk lean and A-P CoP), again averaged across the time window 15-25 % stance. A pooled correlation (all subjects) was performed across subjects from both groups, and separate correlations were conducted with knee OA and healthy groups. Correlation tests were carried out to quantify the strength of the linear relationship between each pair of variables. Guidelines suggested by Hinkle et al. (2003) and Mukaka. (2012) were used to interpret the value of the correlation coefficient and are summarised in Table 4-1 below.

Table 4-1. Correlation level guidelines (Mukaka, 2012, Hinkle et al., 2003)

Coefficient Value	Level of Correlation
0.1 < [r] < 0 .3	*Negligible correlation*
0.3 < [r] < 0.5	*Weak/low correlation*
0.5< [r] < 0.7	*Moderate correlation*
[r] > 0.7	*Strong/high correlation*

4.4 Results

4.4.1 Trunk inclination in people with knee OA

RQ 1A: Do individuals with knee OA walk with an increased inclination of the trunk?

Figure 4-2 shows the ensemble average (across each of the two separate groups, knee OA and healthy) trunk inclination pattern during stance phase. This ensemble profile has the same pattern in both the knee OA and healthy group. The plots show that the knee OA individuals walk with higher forward trunk inclination during whole stance phase. Specifically, in initial contact, the lean was ~ 6 º in knee OA groups, while it was ~ 4 º for the healthy group. However, as noted above, the general pattern was the same, with small decrease in inclination during early stance and a slight increase over midstance. The highest peaks for both groups were at the initial contact and late mid stance. Interestingly, the period of 15-25% appeared to be the period associated with the largest difference across the groups.

During the period 15-25% of stance, the forward trunk inclination was significantly higher ($p<0.01$) in the knee OA group compared to the healthy group, with a mean difference of 3º. Descriptive data for on trunk inclination across the period 15-25% stance are given in Figure 4-2, Figure 4-3 and Table 4-2.

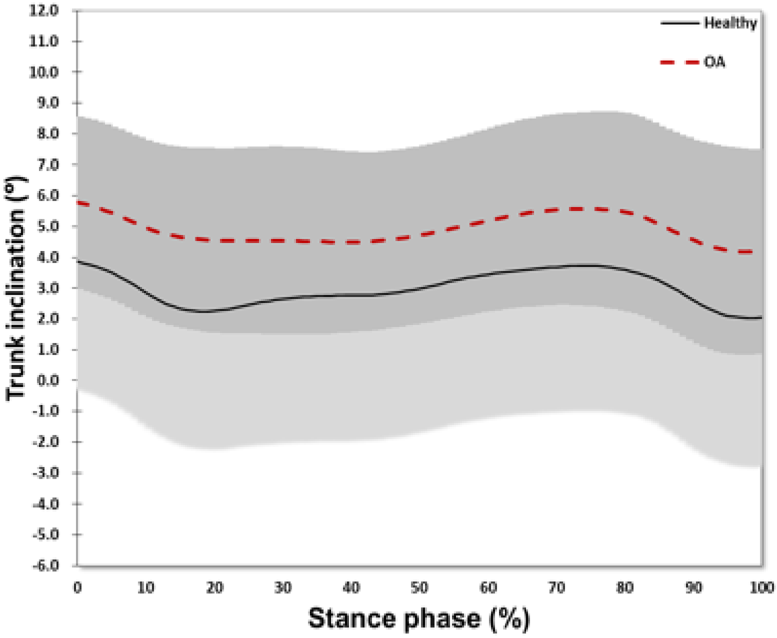


Figure 4-2. Ensemble average and standard deviation of trunk inclination during walking in stance phase for people with knee OA (red dashed) and healthy subjects (black).

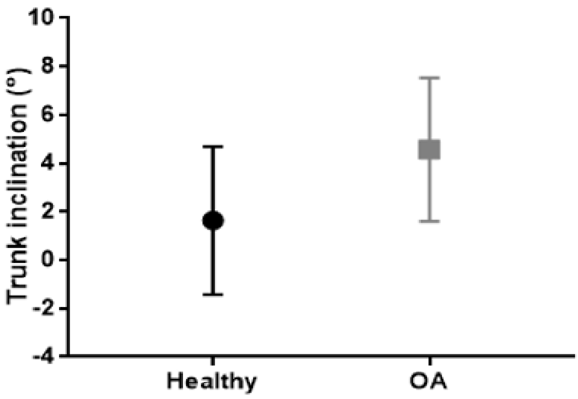


Figure 4-3. Mean (SD) trunk inclination (15-25% of stance) for healthy participants and those with knee osteoarthritis in control shoes

Table 4-2. Mean (SD) forward trunk inclination in specific period of stance phase (15-25%) for healthy and knee OA group.

Group	**Control shoes**
Healthy Mean(SD)	1.6 º (±3)º
Knee OA Mean(SD)	4.6º (±2.9)º
***P* value**	0.002

RQ 1B: Do individuals with knee OA stand with an increased inclination of the trunk?

There was a significant difference between the knee OA and healthy group in terms of standing forward trunk inclination. Specifically, those individuals suffering from knee OA had 1.7º more trunk inclination (p<0.01) in comparison to the healthy group. Descriptive data are shown in Figure 4-4 and Table 4-3.

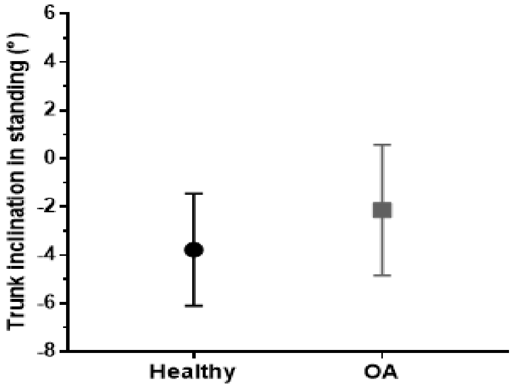


Figure 4-4. Mean (SD) trunk inclination in standing for healthy group and knee OA group. Control shoes.

Table 4-3. Mean (SD) trunk inclination standing in two different shoes for healthy group and those with knee OA.

Grouping	Control shoe
Healthy mean (SD)	(-3.7± 2.3) º
Knee OA mean (SD)	(-2 ± 2.7) º
P value	0.03

RQ 1C: Does trunk inclination in standing correlate with trunk inclination in walking both in a group of individuals with knee OA and also in a healthy cohort?

Based on Hinkle et al. (2003) and Mukaka. (2012) correlation guidelines in Table 4-1, the results showed that there was a positive weak correlation (r =0.42) between trunk inclination in standing and trunk inclination during walking during the specific time period 15-25 % but only when the data from the health and the knee OA participants was combined. In contrast, the correlations calculated separately for each group were not significant (see Table 4-4 and Figure 4-5).

Table 4-4. Correlation between trunk inclination in standing and trunk inclination in walking for group of individuals with knee OA and also in the healthy group.

Type of shoes	Group	r value	p- value
Control shoes	*Knee OA*	**.31**	**.103**
	Healthy	**.38**	**.10**
	Combined	**.42****	**.003**

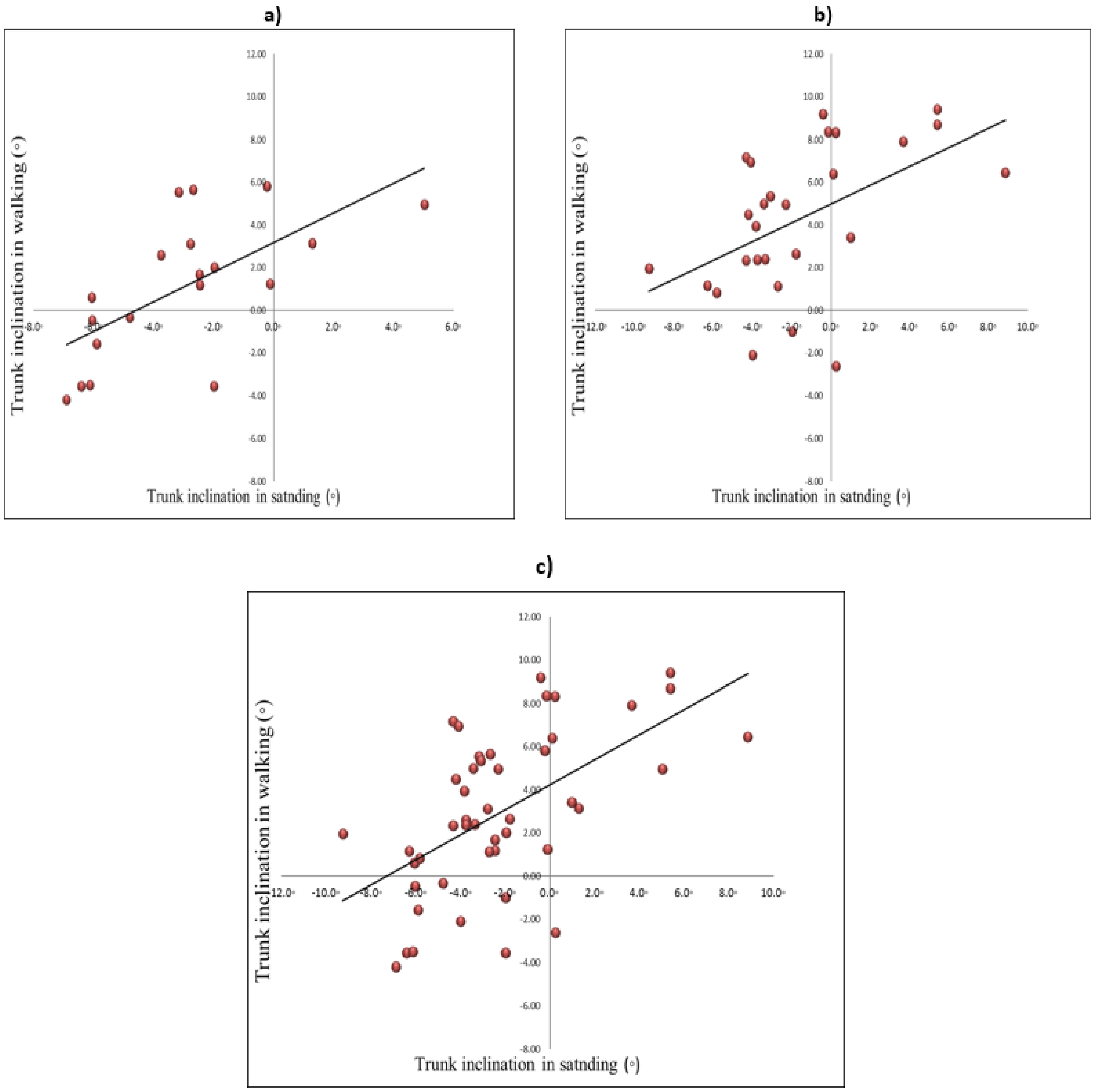


Figure 4-4. Correlation between trunk inclination in standing and walking (15-25 % stance). a) Trunk inclination for healthy group. b) For knee OA participants and c) for combined group.

4.4.2 CoP in people with knee OA

RQ 1D: Is there a difference in CoP between healthy and knee OA subjects?

The plot in Figure 4-6 shows the means (SD) ensemble profile of the anterior-posterior position of the centre of pressure (AP-CoP) for the two groups of participants across the stance phase (%). The plots show a similar pattern for both groups at initial contact and pre-swing with a slight difference during mid stance. Specifically, the AP-CoP was slightly more anterior in the OA group between 20-60% stance with a maximum difference of approximately 1.5 cm occurring around mid-stance. However, at the specific period (15-25%) of stance phase, there were no significant differences ($p>0.05$) in anterior-posterior (A-P) displacement of CoP at knee OA group between the healthy and knee OA participants. Descriptive statistics (mean and standard deviation) are provided in Table 4-5.

Table 4-5. Mean (SD) anterior-posterior displacement of centre of pressure (CoP) in specific period of stance phase (15-25%). For healthy and knee OA group.

Group	**Control shoes**
Healthy mean (SD) in meters (m)	0.004(±.01)
Knee OA mean (SD) in meters (m)	0.006(±.01)
P value	0.61

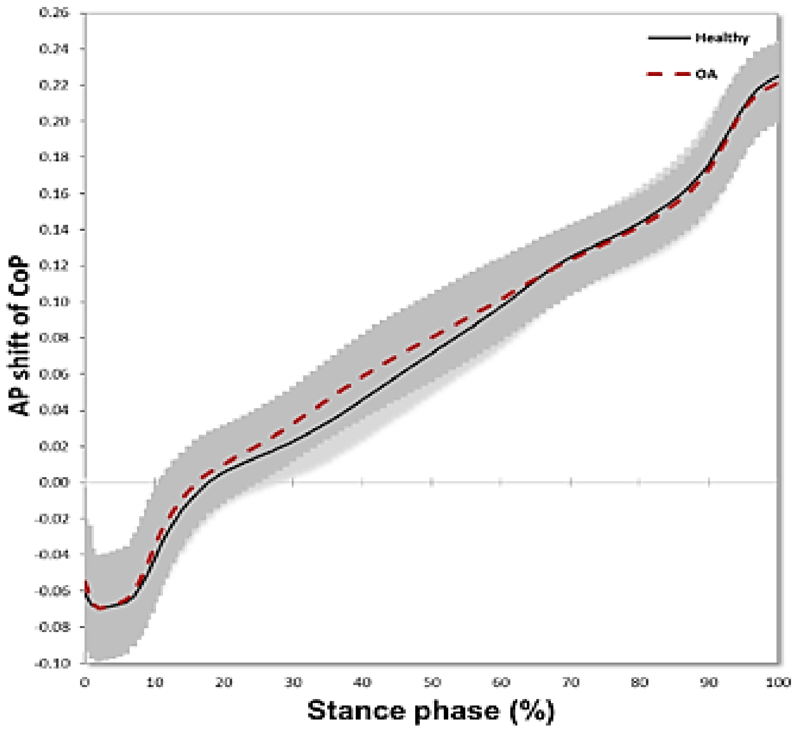


Figure 4-5. Ensemble average anterior posterior (AP) displacement of centre of pressure (Cop) during walking in stance phase for people with knee OA (red dashed) and healthy subjects (black).

RQ 1E: Is there a link between forward trunk inclination and anterior shift of CoP?

A Pearson's coefficient correlation(r) was used to assess the relationship between forward trunk inclination during walking (15-25% stance phase) and CoP A-P displacement. This analysis showed that there was no correlation between these variables for knee OA, healthy and combined groups (Table 4-6).

Table 4-6. Results of correlation between forward trunk inclination and A-P f centre of pressure across 15-25% of stance phase, when wearing control shoes.

Group	r value	p- value
Knee OA	0.078	0.70
Healthy	-0.144	0.55
Combined	0.012	0.93

4.4.3 Sagittal moment

RQ 1F: What are the differences in hip/knee/ankle moments between healthy and knee OA subjects?

Hip moments

The plots below in Figure 4-7 illustrate the ensemble average of the sagittal hip, knee and ankle moments for the two groups during stance phase (%). For the hip moment, these data show only minimal differences between the groups at initial contact, but do illustrate a difference from early to midstance. Specifically, the hip extensor moment was larger for the knee OA group between 20-80% of stance. However, analysis of the period 15-25% of stance showed no significant differences in sagittal hip moment, as reported in Table 4-7. This lack of a statistical difference was likely the results of a large variability across the cohort.

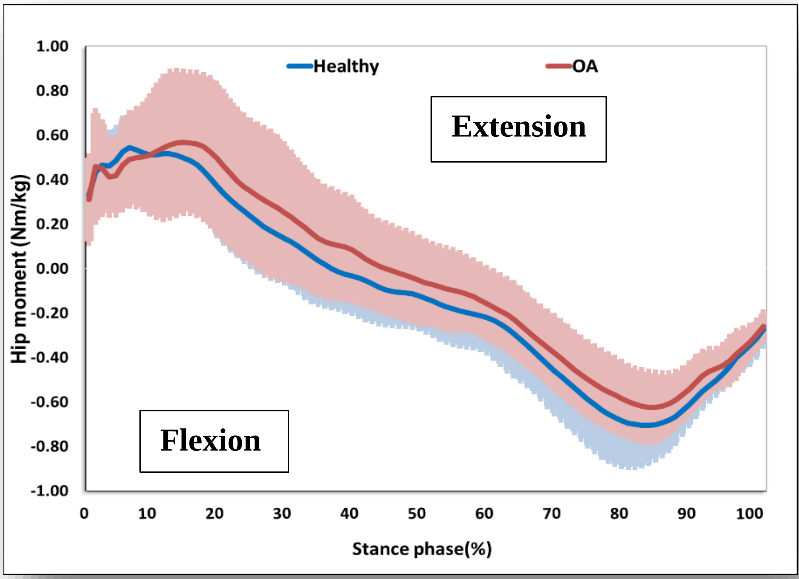


Figure 4-6. Ensemble average of sagittal hip moments for both healthy and knee OA groups.

Knee moment

The sagittal knee moment ensemble average is plotted in Figure 4-8, and shows the difference between knee OA and healthy groups across stance phase. It illustrated that there were only minimal differences between the two groups, with a reduced peak in the knee OA group at around 20% of stance. However, again, there was no significant difference in the average knee moment between 15-25% of stance (Table 4-7).

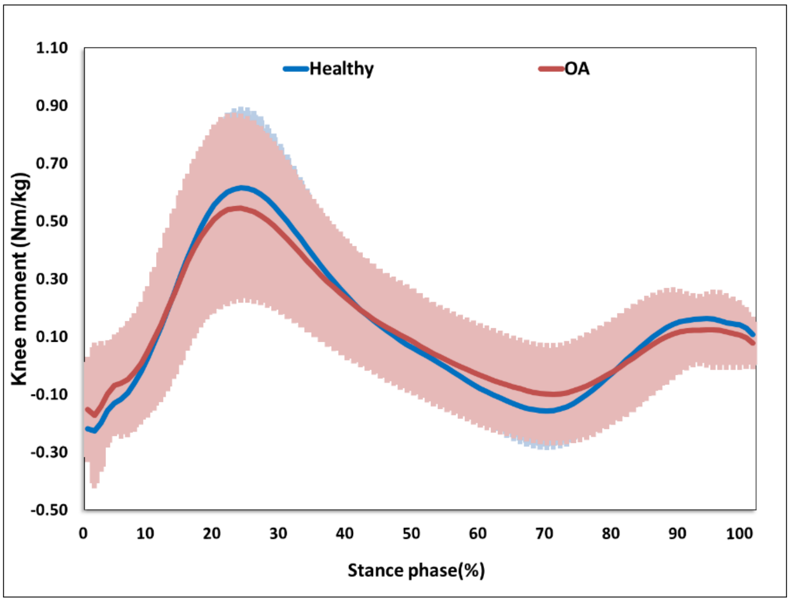


Figure 4-7. Ensemble average of sagittal knee moments for both healthy and knee OA groups.

The ensemble average ankle moment data is shown in Figure 4-9 and again shows only minimal differences between the two groups, with only a slightly high moment in the knee OA group during mid stance. The only other difference was a slightly lower peak plantarflexor moment at the end stance in the knee OA group. Descriptive results for the time period 15-25% stance phase are presented in Table 4-7, and confirm no significant difference between groups over this period.

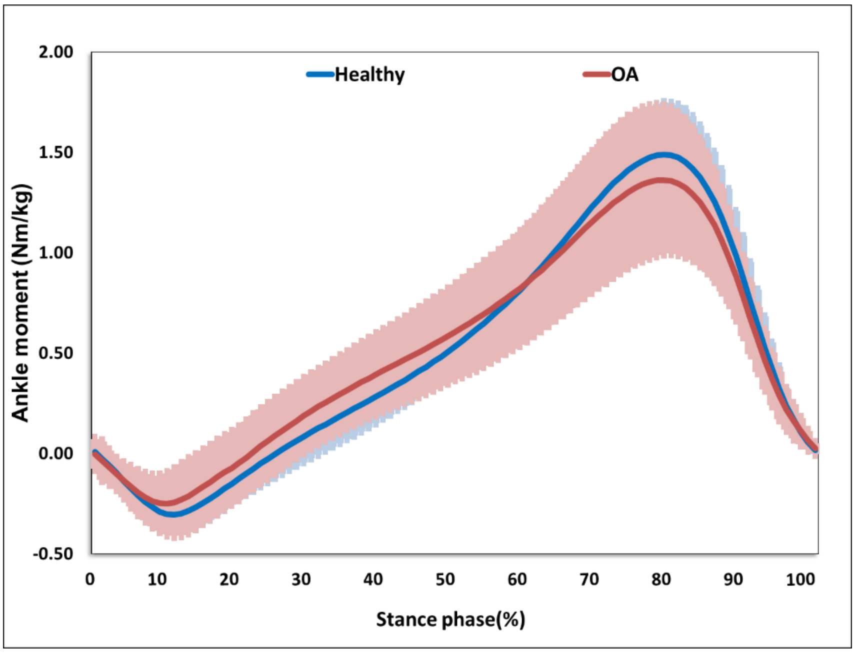


Figure 4-9: Ensemble average of sagittal Ankle moments for both healthy and knee OA groups.

Table 4-7. Mean (SD) sagittal moment of hip, knee and ankle during walking with control shoes for healthy group and individuals with knee OA.

Variable	*Group*	*Control shoes*	
		Mean (SD)	*p-value*
Hip moment	*Healthy*	.34(.20)	**.48**
	OA	.39(.21)	
Knee moment	*Healthy*	.50(.18)	**.96**
	OA	.51(.23)	
Ankle moment	*Healthy*	-.12 (.10)	**.14**
	OA	-.07(.10)	

4.4.4 Differences in muscle activity between healthy and knee OA groups

RQ 1G: What are the differences in hamstring/quadriceps/gastrocnemius muscle activity between healthy and knee OA subjects?

Medial gastrocnemius activity

The plots below show the ensemble average of the normalised medial gastrocnemius activity for both the healthy group and the knee OA group across stance phase (Figure 4-10). This plot illustrates clearly that medial gastrocnemius activity was higher in the knee OA group for almost all of stance phase. Nevertheless, the pattern of activity was similar between the groups with a characteristic rapid rise in activity from 30-65% of stance, followed by a rapid decrease. In the period of interest (15-25% of stance), the knee OA group were observed to walk with ~64% higher muscle activity than the healthy group and this difference was significant ($p<0.05$, Table 4-8).

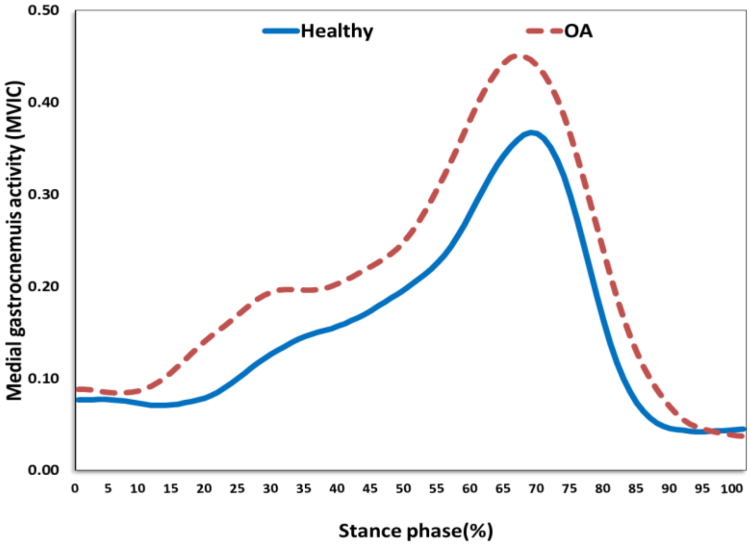


Figure 4-10. The ensemble mean average curves of Medial gastrocnemius activity when wearing control shoes in stance phase for healthy and knee OA group.

Lateral Gastrocnemius activity

Ensemble average profiles for lateral gastrocnemius activity for both groups are shown in Figure 4-11. These profiles were similar to those of the medial gastrocnemius with a distinct peak around 65-75% of stance. Again, there appeared to be increased activation in the knee OA group, however, the difference was more pronounced during the period 0-30% and there appeared to be minimal differences in mid stance. In the period of interest (15-25% of stance), there was significant difference in muscle activity between the two groups ($p<0.05$, Table 4-8), with the knee OA group having almost double the activity of the healthy participants.

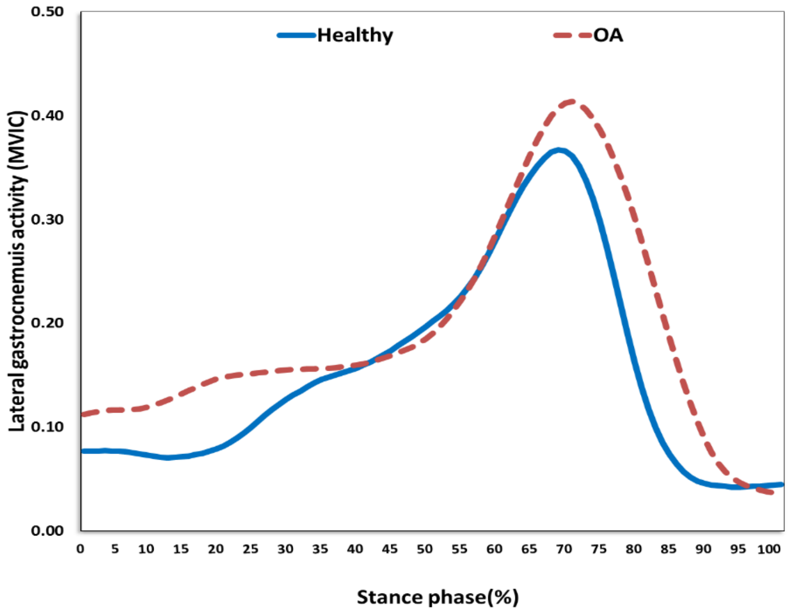


Figure 4-11. The ensemble mean average curves of Medial gastrocnemius activity when wearing control shoes in stance phase for healthy and knee OA group.

Vastus medialis activity

The ensemble average curves in Figure 4-12 illustrate the characteristic pattern of quadriceps activity across the stance phase of walking, with high activity during loading response and early stance. The plot illustrates the differences between the groups, with an overall higher activity in knee OA group. However, these differences were more pronounced around mid stance with minimal differences across the period of interest. The analysis showed, that, although there was a slight difference between the groups across 15-25% stance, this did not reach significance (Table 4-8).

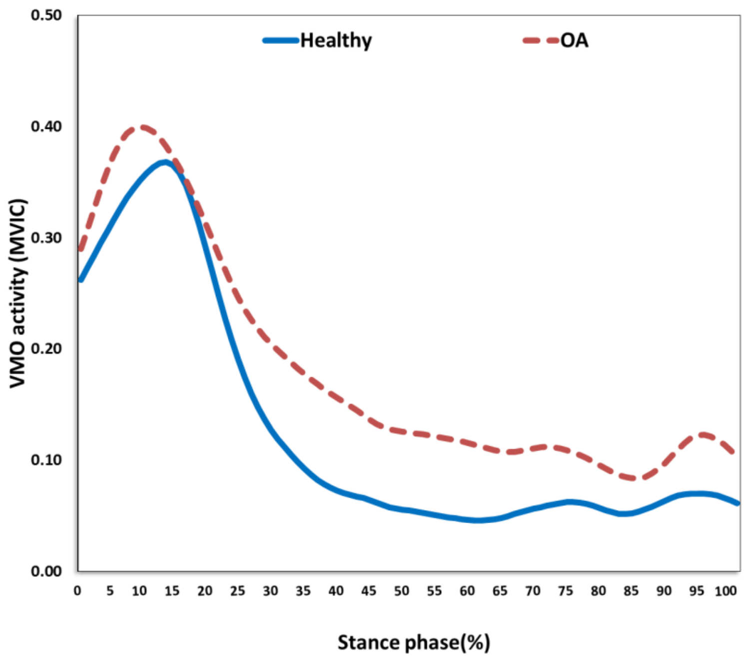


Figure 4-12. The ensemble mean average curves for VM activity during walking with control shoes in stance phase for healthy and knee OA group.

Vastus lateralis activity

The pattern of vastus lateralis activity was similar between the two groups (Figure 4-13). There were very clear differences in vastus lateralis muscle activity between the two groups. However, the ensemble plots show a similar profile of activity for the It was also revealed that the peak value of vastus lateralis activity for both groups was approximately located in loading response and the first half of early mid stance: a similar gait time to peak vastus medialis activity. Interestingly, there was a significant difference in the focused time of stance (15-25%), in which the knee OA group had muscle activation of 57% more than the healthy group, as shown in Table 4-8.

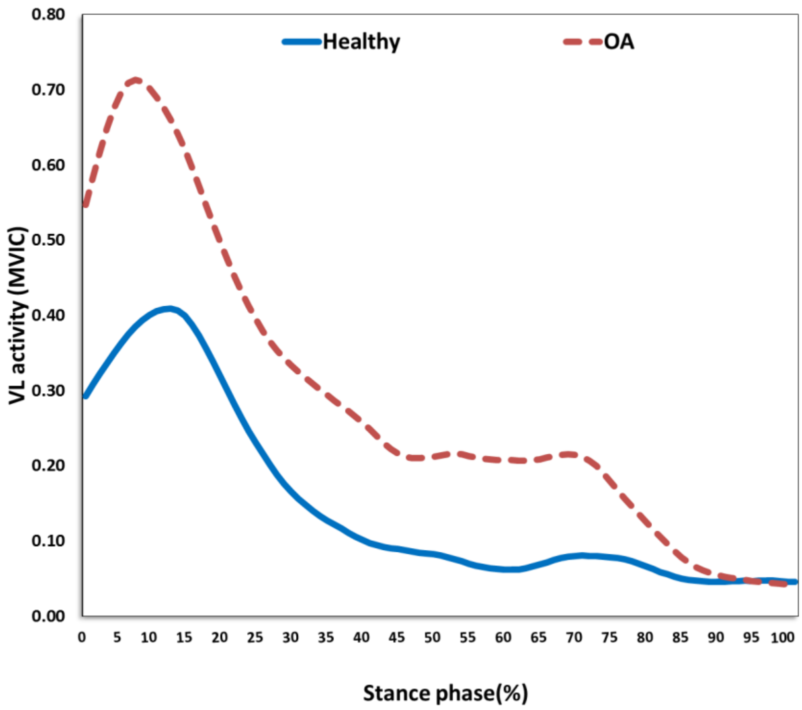


Figure 4-13. The ensemble mean average curves of VL activity when wearing control shoes during stance phase for healthy and knee OA group.

Biceps femoris activity

The ensemble average for the normalized biceps femoris muscle activity (MVIC) for the two groups, knee OA and healthy participants, is presented in Figure 4-14. It was found that maximum activity occurred during the initial contact and loading response for both groups. Then, the activity for both groups decreased in early mid stance. However, in mid stance (early, middle and late mid-stance), those participants with knee OA had approximately the same level of activity as previously. In contrast, for the healthy groups it gradually decreased in mid stance and in the rest of the stance phase. The ensemble average from the plot reflected a clear, significant difference in muscle activity during stance phase. The individuals with knee OA had a higher muscle activity during stance in comparison to the healthy group.

Interestingly, in the specific time of focus for stance phase, 15-25%, there was a significant difference between the groups. Thus, the knee OA group walked with ~54% higher activation than the healthy group. The descriptive data is explored in Table 4-8.

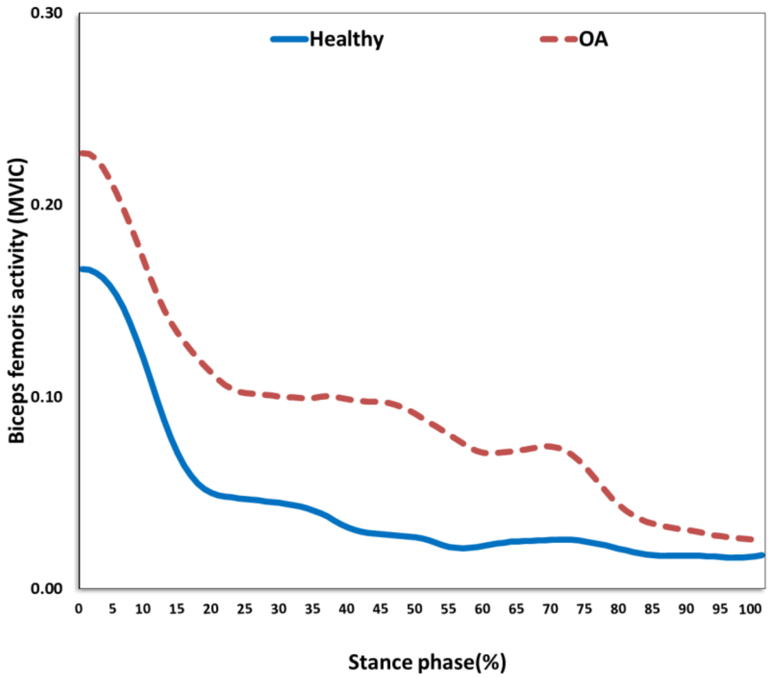


Figure 4-14. The ensemble mean average curves of biceps activity when wearing control shoes during stance phase for healthy and knee OA group.

Semitendinosus activity

The final ensemble average for this research question (RQ 1G) is the normalized semitendinosus muscle activity (MVIC) for both groups, and is presented in Figure 4-15. It follows the same maximum activity pattern as biceps femoris activity, which was during the initial contact and loading response for both groups. However, in the healthy group, this activity decreased fast from initial contact to early mid stance. At two thirds of mid stance (middle and late mid-stance), activity was slightly reduced until reaching minimal activity at pre-swing in the healthy group. However, in the knee OA group, the muscle activity decreased slowly from loading response up to 60% stance, and then recorded minimal activity after that. However, during the whole stance phase, the knee OA group walked with higher semitendinosus muscle activation than the healthy group. Statistically, it is showed in the graph that there was a difference. Specifically, in our time of interest at 15-25% stance, it revealed that there was a statistically significant difference between the groups. This result is illustrated in Table 4-8.

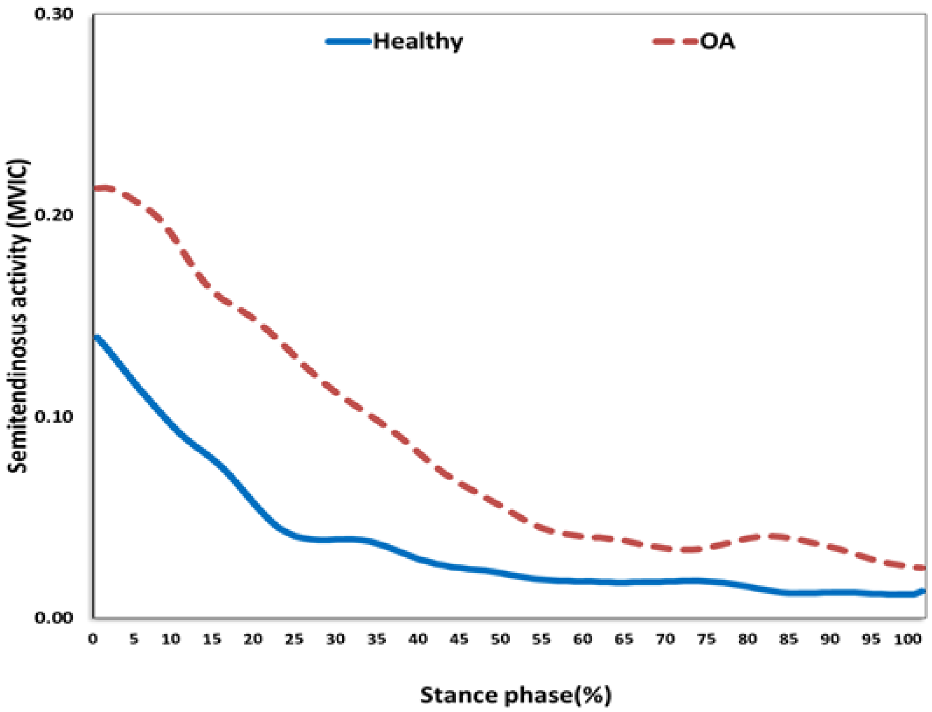


Figure 4-15. The ensemble average of semitendinosus activity when wearing control shoes during stance phase for healthy and knee OA group

Table 4-8. Mean (SD) and P-value of muscle activity of biceps femoris, semitendinosus, vastus medialis, vastus lateralis and medial and lateral gastrocnemius for healthy and knee OA group during walking.

Variable	**Group**	**Control shoe**	
Biceps femoris activity		**Mean (SD)**	***p*-value**
	Healthy	*.07(0.42)*	***.020****
	OA	*.13(.10)*	
Semitendinosus activity	**Healthy**	*.07(.49)*	***.003****
	OA	*.16 (.12)*	
VM activity	**Healthy**	*.34(.27)*	***.814***
	OA	*.36(.21)*	
VL activity	**Healthy**	*.35(.3115)*	***.007****
	OA	*.61(.43)*	
Med gastrocnemius activity	**Healthy**	*.07(.39)*	***.043****
	OA	*.11(.06)*	
Lateral gastrocnemius activity	**Healthy**	*.065(.027)*	***.012****
	OA	.13(.12)	

4.4.5 Medial and lateral muscular co-contraction for healthy and knee OA groups

1. RQ 1H: What are the differences in co-contraction of between healthy and knee OA subjects?

The muscle co-contraction between the biceps femoris and Vastus lateralis was increased significantly (p<.05) in the knee OA group in comparison to the heathy group. Additionally, there was observed a significantly higher (p<.05) muscle co-contraction between lateral gastrocnemius and vastus lateralis in the knee OA group. However, there was an increase, but not a significant one, in the muscle co-contraction between the semitendinosus and vastus medialis, and also between the medial gastrocnemius and vastus medialis (p>.05) in knee OA patients compared to the healthy group.

Table 4-9. Mean (SD) and P-value of muscle co-contraction of biceps femoris and vastus lateralis, semitendinosus and vastus medialis, medial gastrocnemius and vastus medialis and lateral gastrocnemius and vastus lateralis during 15-25% of stance phase in control.

Variable	**Group**	**Control shoes** Mean (SD)	***p*-value**
Vastus lateralis vs. Biceps femoris	Healthy	.41 (.16)	.002*
	OA	.74(.47)	
Vastus medialis vs Semitendinosus	Healthy	.45(.25)	.50
	OA	.52 (.37)	
lateral Gastrocnemius vs. Vastus lateralis	Healthy	.43 (.16)	.003*
	OA	.74 (.46)	
medial Gastrocnemius vs. Vastus medialis	Healthy	.42(.22)	.572
	OA	.47(.34)	

4.5 Discussion

4.5.1 Overview of the results

An overview of the results for healthy and knee OA groups has been provided in Table 4-10. These data show the results of the statistical comparisons for the specific period of stance phase, 15-25%, for all variables measured during walking.

Table 4-10. Summary of results in healthy and knee OA groups

Comparison	Variables	P-value	Null hypothesis	Acceptance / Rejection of null hypothesis	Favour to
Knee OA group Versus Healthy group	Forward trunk inclination during walking	**0.002***	No significant differences between the groups	***Reject***	**OA group**
	Forward trunk inclination during standing	**0.03***		**Reject**	**OA group**
	Cop displacement (A-P)	**0.61**		***Accept***	
	Sagittal hip moment	**0.48**		***Accept***	
	Sagittal knee moment	**0.96**		***Accept***	
	Sagittal ankle moment	**0.14**		***Accept***	
	Biceps femoris activity	**0.020***		***Reject***	**OA group**
	Semitendinosus activity	**0.003***		**Reject**	**OA group**
	VM activity	**0.814**		***Accept***	
	VL activity	**0.007***		***Reject***	**OA group**
	Medial gastro activity	**0.043***		***Reject***	**OA group**
	Lateral gastro activity	**0.012***		***Reject***	**OA group**
	Vastus lateralis vs. Biceps femoris	**0.002***		***Reject***	**OA group**
	Vastus medialis vs Semitendinosus	**0.50**		***Accept***	
	lateral Gastrocnemius vs. Vastus lateralis	**0.003***		***Reject***	**OA group**
	medial Gastrocnemius vs. Vastus medialis	**0.572**		***Accept***	

The main findings of the study reveal significant differences between the OA and control groups in terms of standing and walking trunk inclination. Specifically, the knee OA group stood with an increase in trunk inclination of approximately 2º compared to the healthy group and walked with 3º greater forward trunk inclination than healthy participants. However, there was only a weak correlation between forward lean in standing and in walking, while A-P CoP displacement showed no correlation to walking trunk inclination. Nevertheless, there were clear differences in muscle activations and co-contraction, supporting the idea that people with knee OA use increased muscle activation and co-contraction to walk. The chapter's discussion section will relate these and other findings to the relevant research question, and discuss the implications of the results.

4.4.1 Trunk Inclination in Standing and walking

This study found a significant increase in forward trunk inclination in knee OA during standing when compared to the healthy group, which is consistent with the findings of Turcot et al. (2015). This similarity in the two studies is despite methodological differences. While Turcot et al.'s (2015) knee OA group was limited to patients with severe knee OA, this study tested people with moderate knee OA. The two studies taken together therefore establish a general tendency for knee OA sufferers to stand with a forward inclination of the trunk.

The data on trunk inclination during walking, presented in this thesis, are a genuinely novel contribution to the literature, as no previous studies exist on this parameter. I showed that people with knee OA walk with a clear forward lean. Interestingly, however, when comparing the trunk inclination in walking and standing there was not a strong correlation. This points to the possibility of a different mechanism underlying the forward lean in standing and the forward lean in walking. Whereas Turcot et al. (2015) suggested that standing forward lean, along with increased sagittal flexion in the joints of the lower limb, in people with knee OA may be an attempt to move the centre of mass forward to aid with balance, it is possible that the forward lean in walking may be due to muscular restriction. The potential mechanisms behind increased trunk inclination when walking will be discussed later in this section, but first, the findings for A-P CoP will be discussed.

The investigation of anterior-posterior displacement of CoP did not give statistically significant results. However, a difference between the healthy and knee OA groups was observed at mid-stance, with the OA group's CoP being shifted forwards from 15% of stance. This change of CoP reached approximately 1.5cm at around midstance, and although not statistically significant does suggest that people with OA have a tendency to shift load on the forefoot during mid stance. This idea is consistent with the findings of Saito et al. (2013) who also observed a shift in the CoP from the heel to the mid foot during early-mid stance.

The data did not show any meaningful correlation between forward trunk inclination and A-P CoP. This is finding is surprising, and does not support the idea that increasing forward lean will shift the centre of mass anteriorly and this will lead to a corresponding anterior shift in the CoP. It is possible that this lack of a correlation points to the idea that increases in forward trunk lean are compensated for by a series of small subtle changes at each joint, rather than a simple, and proportionate, shift in the CoM. The lack of a correlation is also not consistent with the idea that increasing trunk lean is a strategy used to anterior shift the CoP anteriorly to aid balance as suggested by Turcot et al. (2015) to explain their observations in standing.

It is important to use the finding so this study to speculate on the underlying cause of increased trunk inclination in people with knee OA. I suggest a number of mechanisms. The first mechanism relates to tightness of the hip flexor muscle which can lead to compensatory hip movement (Kagaya et al., 2003) . In older people, it is common to develop muscle imbalances around the hip, which are characterised by hip flexor muscles tightness and gluteal muscle weakness. As the tight muscle becomes shorter in length, this will change the pelvis position and create a corresponding change in the trunk position (see Figure 4-16). There is some evidence to support this idea, as Sato and Maitland (2008) suggest tight hip flexor muscles as an underlying factor in forward lean while standing and walking. While their study focused on females from 46-79, and not knee OA sufferers, the mechanism may be applicable here. This idea is also supported by a simulation of walking with joint contractures carried out by Kagaya et al. (2003), who found that in a simulated model, introducing hip flexion contracture led to a forward leaning of the trunk. However, this idea has not previously been discussed in relation to knee OA, and so it motivates further research to investigate hip flexor tightness in those with knee OA.

The second potential mechanism is a strategy in which people with knee OA adopt a forward-leaning posture because of balance issues. Specifically, I suggest that the forward lean may be a strategy to shift the CoP anteriorly in order to reduce the risk of falling backwards. Previous studies demonstrate balance issues for those with knee OA, shown through an increased fall rate (Levinger, 2011) and greater postural sway while standing on one leg (Tarragon, 2009). This could relate to the alterations observed in muscle activation patterns while walking (Duffell, 2013), and contribute to explanations of forward lean. However, as explained above, the lack of a correlation between trunk inclination and CoP does not support the idea that the alteration in trunk lean was a direct strategy to shift CoP and therefore influence balance. A third possible mechanism is that forward lean is adopted due to a feeling of load on the knee joint. However, no significant changes were noted in sagittal knee joint moment in this study, a finding which is consistent with the observations of Baliunas et al. (2002). In conclusion, given the data presented in the section above, I suggest that the most likely explanation of increased forward lean in the knee OA group in related to tightness of the hip flexor muscles.

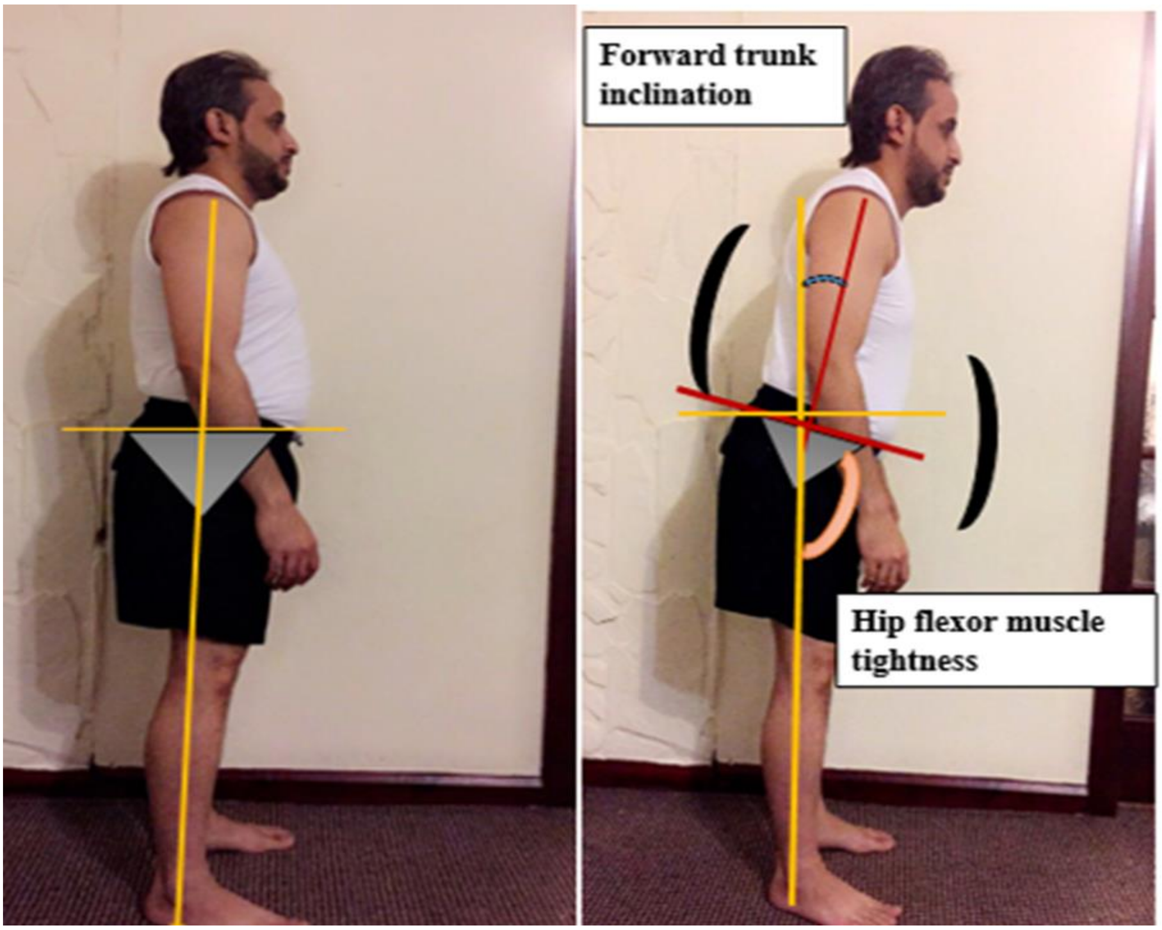
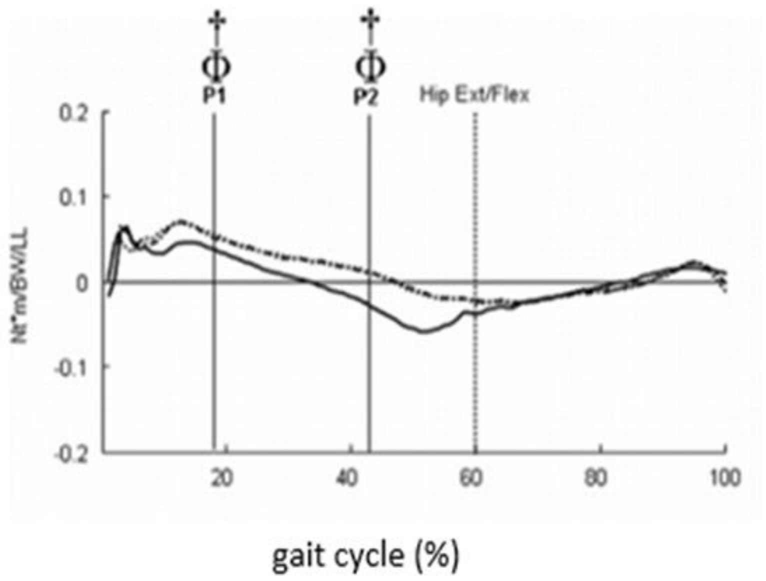


Figure 4-16. Upright posture versus forward trunk inclination due to hip flexor muscle tightness. The first picture shows a normal upright posture with normal inclination of the trunk. In the second, tight hip flexor muscles cause the trunk to incline forward.

In light of the discussion above, clinically, strength exercises for the gluteus maximus muscle could help to decrease forward trunk lean and decrease the external extension moment of the hip joint. Also, it may be useful to increase the strength of the spinal muscle (the back muscle of the trunk). Further, the problem could also be treated using a specifically-designed footwear intervention (discussed in Chapter 6), and through stretching of the hip flexor muscle (Falconer et al., 1992). However, further investigations are required before a particular clinical intervention can be recommended.

4.4.2 Differences in moment between knee OA and healthy groups

The study explored sagittal lower limb joint moments and found minimal differences between the groups. Nevertheless, it is interesting to compare the sagittal hip data in this study with that observed by Liu et al. (2014). In this study, while the difference between the two groups was not statistically significant, mean hip moment over the period of interest was up to 15% higher in the knee OA group compared with the healthy sample. This difference was at its greatest from the first peak, in early mid-stance, and continued throughout most of the gait cycle. Liu et al. (2014) found a significantly higher extensor moment in the OA group in comparison to a group of healthy controls (see Figure 4-17). Visual inspection of the two plots (shown below) show a similar difference in the overall pattern and therefore, despite the lack of statistical significance in the current study, a consistency in the findings. It is possible that the larger effect observed by Liu et al. (2014) was the result of more forward lean in the group tested by Liu et al. (2014). However, this gait characteristic was not reported and so it is not possible to make any definite conclusions about the underlying reason for the differences.



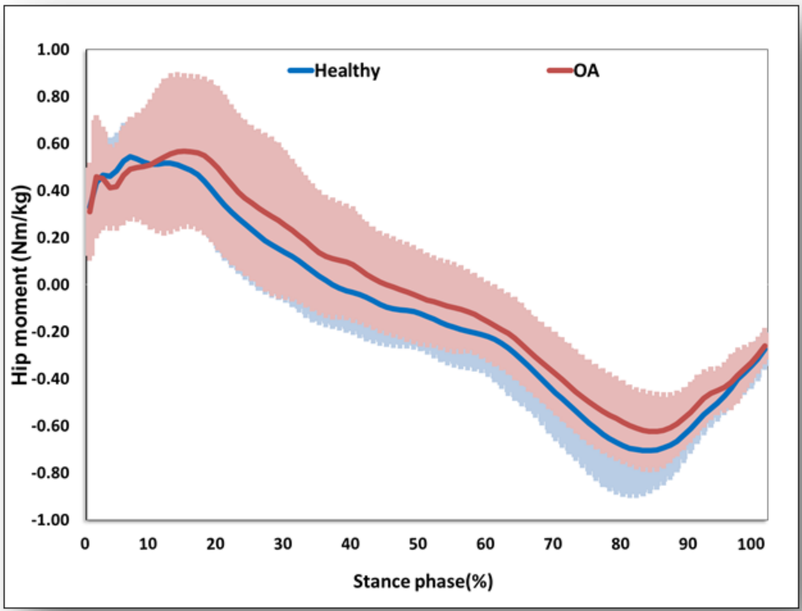


Figure 4-17. Hip extension/flexion moments in Liu et al. (2014) (above) and in the current study (below). Hip flexion/extension moments are shown across the gait cycle in the first graph and across the stance phase only in the second graph, with 1-100 % in the second graph equal to 1-60% in the first. A similar pattern is seen in the two studies, with increased values for OA groups.

In another study Astephen et al. (2008) found reduced early stance hip extension moment in a group with severe OA, which further supports the overall finding of this study, shown in the figure above. However, it is important to note that the lack of a significant difference in the current study may be due to the focus on the specific period of 15-25% of stance. In Figure 4-17 above, there appears to be a more pronounced difference in midstance and this many have been significant. Nevertheless, we chose to focus on the period 15-25% of stance because this corresponds to the period of peak loading at the knee (Brandon et al., 2014; Sritharan et al., 2016), it was felt that focusing on multiple time periods would have increased the number of statistical tests and therefore the probability of type 1 error.

This study did not find any significant differences in the sagittal plane knee moment between the knee OA group and healthy controls, with an increase of just 2% for the OA group. This finding contrasts with a number of previous studies (Kaufman et al., 2001; Astephen et al., 2008a; Sritharan et al., 2016; Mündermann et al., 2005; Liu et al., 2014), which showed a statistically significant reduction in knee moments in the sagittal plane in participants with knee OA. However, these studies tended to focus on the peak knee moment and not the mean knee moments across a specific period of the gait cycle (as was analysed in this study). Nevertheless, Baliunas et al. (2002) observed similar trends to our data, finding no significant differences in peak knee extension/flexion moments or in early midstance knee extension/flexion. As explained above, I chose to focus on the period 15-25% it was felt to be the period most appropriate to use in order to understand joint loading.

The ankle moment across the period 15-25% showed an increase in the group with knee OA, however, this was not statistically significant. These findings match those of Liu et al. (2014), who also report an increased sagittal ankle moment, which was significant in a group with severe OA group but not in a moderate group. This moderate group are more characteristic of this study and so there appears to be a consistency in the findings. The observation of no increased ankle moment matches the findings of no significant change in the CoP position and, as explained earlier, suggests that the increases in trunk inclination characteristic of the people with knee OA are not accompanied by an anterior shift of the CoM and therefore the CoP.

4.4.3 EMG

The next element of the study was measurement of mean muscle activity for 6 muscles involved in walking: the biceps femoris and semitendinosus from the hamstrings; the Vastus medialis and vastus lateralis of the quadriceps; and the medial and lateral gastrocnemius muscles. In general, the results for the OA group show raised activity levels in the 15-25% period across 5 of the 6 muscles studied, with the exception being the Vastus medialis. These finding are similar to those of previous studies comparing muscle activation between knee OA and healthy groups (Childs et al., 2004; Hortobagyi et al., 2005; Zeni et al., 2010; Hodges et al., 2016). In interpreting these results, raised EMG levels can be understood to link to increased loading on the joints, based on the modelling papers used for the study (Brandon et al., 2014; Hodges, 2016). The results therefore imply increased knee loads for the OA group.

The vastus medialis and lateralis were measured for the quadriceps group. The findings showed that there was high activation overall in the group with knee OA, but not a significant difference over the 15-25% period for the VM. Interestingly this period corresponded to the highest VM activity across both groups. This lack of a significant difference in VM activity is surprising as it does not match the findings of previous studies (e.g. Sharma et al., 2017). However, the overall pattern of VM activity showed a clear increase in the knee OA group, with the similarity, during the period of interest being uncharacteristic of the general profile.

Nevertheless, although these results are similar to those reported by Selistre (2017), it is not clear why they do not match the findings of other research (Sharma et al., 2017).

The findings for the vastus lateralis reveal significantly higher activity in the OA group during the stance phase. These findings are in line with Zeni et al. (2010), who found significantly greater quadriceps activation for moderate OA subjects over healthy controls when walking at a fixed, fast walking pace, with results at a self-selected speed also showing an increase but not reaching statistical significance. The findings of Hortobagyi et al. (2005), Hodges et al. (2016) and Childs et al. (2004) also support those of the current study and provide general support for increased quadriceps muscle activation in knee OA.

For the hamstrings, significantly higher biceps femoris activity was found in the OA group, and this muscle remained active through midstance, compared to healthy controls for whom BF activity decreased. Semitendinosus activity was also higher and maintained for longer compared to a rapid decrease in the healthy group through the stance phase. These results are in line with those of Hortobagyi et al. (2005), Zeni et al. (2010) and Sharma et al. (2017), who also found increased activity. Taken together these findings support the idea that increased hamstring activity is a clear gait characteristic of people with knee OA. In the next chapter, I explore whether this increase is associated with an increased trunk lean.

As with the hamstrings and quadriceps, increased gastrocnemius muscle activity was also seen in the OA group while walking. In fact, for both the medial and lateral gastrocnemius muscles, there was significantly higher activity in early midstance for the period 15-25%, continuing until 65-75% of stance. Previous assessments of gastrocnemius activity in knee OA have made similar findings (Schmitt & Rudolph, 2007; Childs et al., 2007; Sritharan et al., 2016a) and again, my data support the idea of generalised increased activity in posterior calf muscles as a gait characteristic of people with knee OA.

4.4.4 Co-contraction

Co-contraction was studied during walking for the muscles of the knee, including the pairs: vastus lateralis vs. biceps femoris; vastus medialis vs. semitendinosus; lateral gastrocnemius vs. vastus lateralis; and medial gastrocnemius vs. vastus medialis. For all pairs studied, there was increased co-contraction shown in the OA group compared to healthy controls, but only the VL-BF and LG-VL findings were statistically significant. The increase in BF-VL co-contraction was also reported by Sharma et al. (2007), who link increased co-contraction activity between quadriceps and hamstrings with increased hamstring activity during walking. Zeni et al. (2010) also find that increased quadriceps-hamstring co-contraction is a strategy employed during gait by those with moderate OA.

As explained in the literature review, increased muscle co-contraction will increase the compressive loading at the knee joint. Although some authors have suggested that increased co-contraction is employed as a neuromuscular approach to promote unloading of the medial compartment Andriacchi (1994), there is no strong evidence to validate this idea. Nevertheless, there is evidence that increased co-contraction will accelerate cartilage degeneration (Hodges et al., 2016) and also increase the likelihood of progression to total knee replacement (Hubley-Kozey et al., 2013). One of the ideas explored in this thesis is that increased co-contraction could be the result of increased trunk inclination. In this chapter, I have shown that people with knee OA walk with increased trunk lean and increased co-contraction. The possible links between these two phenomena will be explored further in the next chapter of the thesis.

4.4.5 Limitations

The measurement of trunk lean is problematic. Specifically, it is challenging because of the complex and flexible structure of the spine in comparison to the simpler and more rigid structures of the lower limbs, such as the femur. In this study, trunk lean was quantified by defining a single thoracic segment which was tracked using three markers attached to the IJ, T2 and T8, and defined using four markers attached bilaterally on the acromiums and the greater trochanters. This thoracic segment was tracked against the laboratory reference frame and this study focused specifically on sagittal movements. This approach was adopted given the recommendations of a previous study (Armand et al., 2014), showing that this method gave the most valid and reliable data, as the set of markers with the lowest error when tested from a range of possible markers. However, limitations remain because minor anatomical differences in the shape and position of the anatomical structure used to define and track the segment could lead to differences in trunk angle. This could lead to uncertainty in quantifying trunk inclination. Nevertheless, the data did demonstrate a clear difference in trunk lean between people with knee OA and healthy control subjects.

The study took its focal point as the 15-25% period of stance. This decision made it difficult in some cases to compare the findings with previous work. However, it was important to select a time period which reflected the period of peak loading as ultimately, it was important to explore how trunk lean could impact on the loads at the knee joint. The decision to focus on 15-25% stance was motivated by previous studies that modelled the effect of increasing co-contraction (Brandon et al., 2014; Sritharan et al., 2016). Data from these studies suggest that this period contained a peak in medial and lateral loading which increased markedly when co-contraction was increased (See section 2.4.3 for more details). The focus on a single time window minimised the chance of type 1 errors which would have been more likely if I had chosen multiple periods of the gait cycle to analyse. However, this narrow focus means that the statistical analysis did not capture changes in muscle activation across the whole of stance phase. Nevertheless, the data presented on co-contraction is consistent with previous research which looked at alternative time windows and this may suggest that the findings are not overly sensitive to the precise choice of time period.

Unlike some other studies in this area (e.g. Liu et al., 2014; Astephen et al., 2008), the current study used only one group of subjects with knee OA, rather than separating subjects by severity. This was done to allow a clear focus on the main issues, to ensure appropriate statistical power and to avoid possible logistical problems which might have resulted from recruiting people in different stages of knee OA. The group chosen represents moderate OA, and it is therefore not clear whether different results would be seen with a group in the later stages of disease progression.

4.4.6 Conclusions

In summarising the main findings for the first study of this thesis, it is important to stress the novel findings related to trunk inclination, and anterior-posterior position of the centre of pressure (AP-CoP) during gait. However, the data presented in this thesis also supports previous studies. In particular, the findings for muscle activity, co-contraction and sagittal joint moments in general agree with previous work, although some findings did not show the difference previously shown. I have suggested that this might be due to the focus on a specific period of the gait cycle and/or a single OA group of only moderate severity.

The main original contribution of the study was to demonstrate a significantly increased tendency for forward trunk inclination in knee OA subjects while walking. Previous work has identified forward lean in OA groups while standing, and this study found the same, but these two findings did not correlate with each other in the study, meaning that the underlying mechanisms for this lean may differ. With the lack of a finding of a significant alteration in

A-P CoP during walking, I suggest that the reason for increased trunk lean may related to increased hip flexor tightness. However, further work is required to fully explore this idea. Forward trunk lean while walking may link to and partly explain a range of other mechanical alterations in the gait of knee OA patients. These ideas are explored in more detail in the next chapter.

Chapter 5 - Study two: The relationship between trunk inclination and joint moments/muscular co-contraction

Overarching aim

To investigate the potential link between trunk inclination and biomechanical variables, related to joint loading: lower limb joint moments, muscle activity and co-contraction.

5.1 Overview of the study

This study investigates the possible link between trunk inclination and lower limb moments/muscle activation patterns in the sagittal plane while walking. There has been very little study of the effects of upper body position on lower limb biomechanics and therefore this is novel question, especially in the field of knee OA research. Turcot et al. (2015) found that subjects with knee osteoarthritis show an increased forward trunk inclination during stance, and study one in this thesis both supports this conclusion and also identifies an increased lean in people with knee OA while walking. Interestingly, previous research has shown that trunk inclination in healthy subjects is associated with increases in hip moments during the early and mid-stance phase of walking (Leteneur et al., 2009), and in study one for this thesis, 15% higher hip moments were seen in the knee OA group during this period, with increased hip moment in knee OA also identified by Liu et al. (2014) in mid-stance. This leads to the need to establish whether these variations in joint moments correlate to forward trunk inclination in walking. In addition, Leteneur et al. (2009), in investigating differences in moments as associated with trunk inclination, also found an association between forward trunk lean and a minor decrease in knee moment, as well as with a small increase in ankle moment. The patterns reported in the previous chapter are similar to those reported for individuals with knee OA in previous studies

(Zeni and Higginson, 2011). However, to date, there has been no previous research investigating whether these patterns are intrinsically linked to variations in forward trunk inclination.

It is likely that the increases seen in hip moment are associated with increased levels of hamstring activity, as the major hip extensor muscles. Furthermore, it is possible that these increases in hamstring activation may lead to increased co-contraction, as increases in the knee flexor moment (resulting from increased hamstring activity) may need to be balanced by changes in quadriceps activity to maintain knee joint moment. Increased hamstring-quadriceps co-contraction has been widely reported in knee OA (Zeni et al., 2010; Childs et al., 2004; Hortobagyi et al., 2005) and is linked by Andriacchi (1994) to increased knee joint loading in the medial compartment. Steultjens et al. (2006) argue that higher rates of muscle activity in knee OA are a positive defence mechanism against passive joint instability and are needed to maintain walking ability, while Lewek et al. (2005) argue that interventions to reduce co-contraction are necessary to relieve joint loading and slow progression of the disease. In light of this debate, it is interesting to investigate forward trunk inclination as a potential factor in the reported increase in lower limb muscle activation and co-contraction in knee OA. This will provide some insight into the mechanisms which could underlie co-contraction in people with knee OA.

The peak compressive knee joint loading has been identified by modelling studies as occurring at between 15% and 25% of the stance phase (Sritharan et al., 2016; Brandon et al., 2014). Interestingly, Leteneur et al. (2009) found significant increases in hip extensor moment during this phase for individuals who have a forward trunk inclination while walking, with a corresponding delayed transition from extensor to flexor moment. This period was investigated in the previous study for this thesis and will also be used in this chapter. The main findings from study one include a significant difference between knee OA and healthy participants in forward trunk inclination.

In the previous section, I presented data showing an increase in forward trunk inclination in people with knee OA. I then discussed a scenario in which, as the trunk inclines further forward, possible anterior shifting of the centre of mass is compensated for by movement at the knee and ankle joints with a corresponding minimal change in the centre of pressure, remaining static. This idea is supported by the findings, which showed only a minimal shift in CoP. Nevertheless, with this model, the hip joint centre would move anteriorly, increasing hip extensor moment with little change in GRFV, possibly reducing knee extensor moment with anterior movement of the GRFV, and causing little change to dorsiflexion and plantar flexion moment at the ankle. These are similar changes to those seen in walking in knee OA, and these changes, their impact on muscle activity, and their correlation to trunk inclination, are therefore investigated here.

The second study of this PhD thesis consists of three research questions, which are given in the next section. These questions build from the results of the first study to focus on forward trunk inclination during walking in knee OA and healthy subjects and investigate potential links between this and lower limb joint moments, muscle activity and co-contraction.

Therefore, the study addresses the following questions:

5.2 Research questions

RQ 2A: What is the relationship between trunk inclination and hip/knee/ankle moments in people with knee OA and also in healthy control subjects?

RQ 2B: What is the relationship between trunk inclination and hamstring/quadriceps/gastrocnemius activity in people with knee OA and also in healthy control subjects?

RQ 2C: What is the relationship between trunk inclination and co-contraction in people with knee OA and also in healthy control subjects?

5.3 Methodology

A detailed description of the procedures carried out for each of the studies is given in the methodology chapter. The study was carried out with subjects wearing control shoes only. All data collection was conducted in the Allerton Building Gait Laboratory located within the University.

5.3.1 Sample and population

The data for the second study were collected during one visit across two groups; healthy (n=20) and knee OA (n=27) subjects, across both genders. Further details of the specific inclusion/exclusion criteria are provided in Section 3.2 of the methods section.

5.3.2 Derivation of outcome measures and statistical methods:

This study was designed to determine the association between forward trunk inclination over 15-25% stance and lower limb joint moments, muscle activity and co-contraction. The methods section explains how specific joint kinematics and moments related to the research questions were derived and also details the post-processing which was applied to the raw EMG data. In the section below, I have explained in detail how the specific outcomes (relating to each research question) were derived from the kinematic, moments and EMG data.

RQ 2. What is the relationship between trunk inclination and hip/knee/ankle moments in people with knee OA and also in healthy control subjects?

Mean trunk lean and sagittal hip knee and ankle moments (over 15-25% stance) were the outcomes used to address this research question. Trunk inclination was calculated as set out in the methods chapter, and in the previous chapter for RQ 1A, Sagittal moments for hip, knee and ankle were then calculated as, explained in RQ 1F, Section 4.3.2.

RQ 2b: What is the relationship between trunk inclination and hamstring/quadriceps/gastrocnemius activity in people with knee OA and also in healthy control subjects?

Trunk inclination in walking was defined as explained above for RQ 2A. EMG profiles were then created for the two hamstring muscles (biceps femoris and semitendinosus), the two quadriceps muscles (vastus medialis oblique and vastus lateralis) and the two gastrocnemius muscles (medial and lateral gastrocnemius), as explained in Section 4.3.2 for RQ 1G. The profiles from each muscle in a group (e.g. biceps femoris and semitendinosus for hamstrings) were also summed to give a value for combined hamstrings, combined quadriceps and combined gastrocnemius muscles.

RQ 2c: What is the relationship between trunk inclination and co-contraction in people with knee OA and also in healthy control subjects?

To address this question, I examined the relationships between trunk inclination while walking and muscle co-contraction across the period 15-25% of stance. The outcomes explained in RQ 1H in the previous chapter (Section 4.3.2) relating to EMG co- contraction in walking and RQ 1A, relating to trunk inclination in walking, were used to address this research question.

In the section below, I have provided a detailed explanation of statistical methods used to address each question using the outcomes detailed above.

5.3.3 Statistical analysis

All statistical tests for this study, relating to research questions RQ 2A, RQ 2B and RQ 2C, were performed using the statistical package for social studies (SPSS) version 23 for Windows. This study involved one independent variable, the tested group; and evaluated the correlation with the tested dependent variables. Before the results were analysed, data screening was carried out for normality and homogeneity of variance assumptions, to allow parametric calculations of analysis of difference. No outliers were found when a box and whiskers boxplot was carried out for each tested variable. Descriptive analysis using histograms with normal distribution curves revealed normal distribution of data each variable, and normality tests did not suggest significant differences between the distribution of the tested sample's raw data and normally-distributed population data, using the same means as those of the tested variables ($p>0.05$) and evaluated using the Shapiro-Wilk's test. Homogeneity of variances was identified using Levene's test for equality of variances testing ($p>0.05$) for each dependent variable. Thus, the results did not violate the parametric assumption, and parametric data analysis was possible. Data are given in the form of mean ± standard deviation (SD) in the descriptive statistics. Correlation tests were carried out to quantify the strength of the linear relationship between each pair of variables.

In order to investigate the relationship between forward trunk inclination and the other outcomes defined above (moments, each muscle's EMG profile/each combined profile activity and co- contraction) across the time window 15-25 % stance, associations were calculated using a Pearson's correlation coefficient (r). A pooled correlation was performed across subjects from both groups, and separate correlations were conducted with knee OA and healthy groups. Each Pearson's correlation was run to assess the relationship between trunk inclination and other dependent variables, along a range of r values which were considered to indicate a weak, moderate or strong correlation. Guidelines suggested by Hinkle et al. (2003) and Mukaka. (2012) were used to interpret the value of the correlation coefficient and are summarised in Table 5-1 below

Table 5-1. Correlation level guidelines (Mukaka, 2012, Hinkle et al., 2003)

Coefficient Value	Level of Correlation
0.1 < [r] < 0 .3	*Negligible correlation*
0.3 < [r] < 0.5	*Weak/low correlation*
0.5< [r] < 0.7	*Moderate correlation*
[r] > 0.7	*Strong/high correlation*

5.4 Results

5.4.1 Relationship between trunk inclination and hip/knee/ankle moments

Analysis showed that there was a weak-moderate positive correlation between trunk inclination in walking and hip moment. Specifically, the knee OA and healthy group showed moderate positive correlation, while the combined groups showed a weak positive correlation (see Table 5-2). Figures 5-1 and 5-2 show the correlation between trunk inclination and the hip moment for the healthy and knee OA group respectively. No meaningful correlations were observed between trunk inclination and sagittal knee or ankle moment (Table 5-2).

Table 5-2. Correlation between trunk inclination and hip/knee/ankle moments averaged across 15-25% stance.

Variable	Group	Control shoes	
		R	*P*
Relationship between trunk inclination and hip moment	*Healthy*	.548**	.015
	OA	.545**	.022
	Combined	.495*	.001
Relationship between trunk inclination and knee moment	*Healthy*	.197	.629
	OA	.082	.691
	Combined	.102	.510
Relationship between trunk inclination ankle moment	*Healthy*	-.119	.629
	OA	.175	.391
	Combined	.136	.374

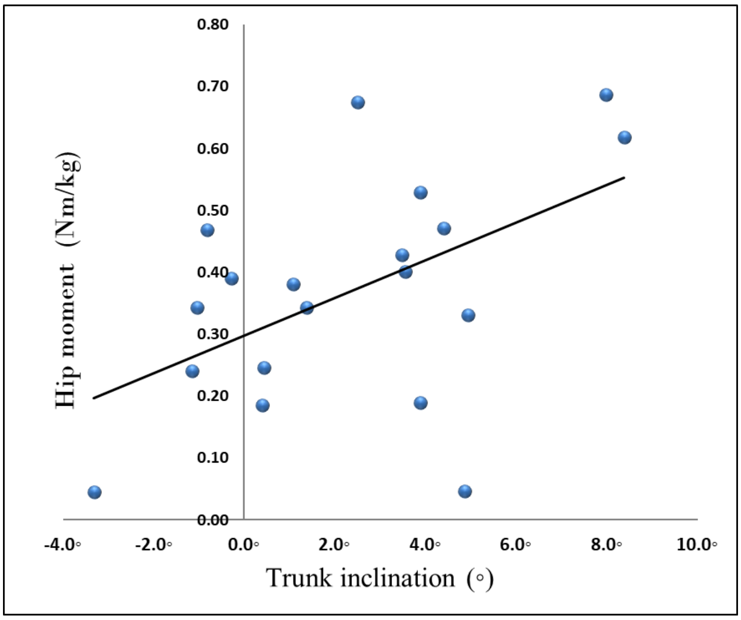


Figure 5-1. Correlation between trunk inclination and hip moments averaged across a specific period of stance phase (15-25%) during walking for healthy group

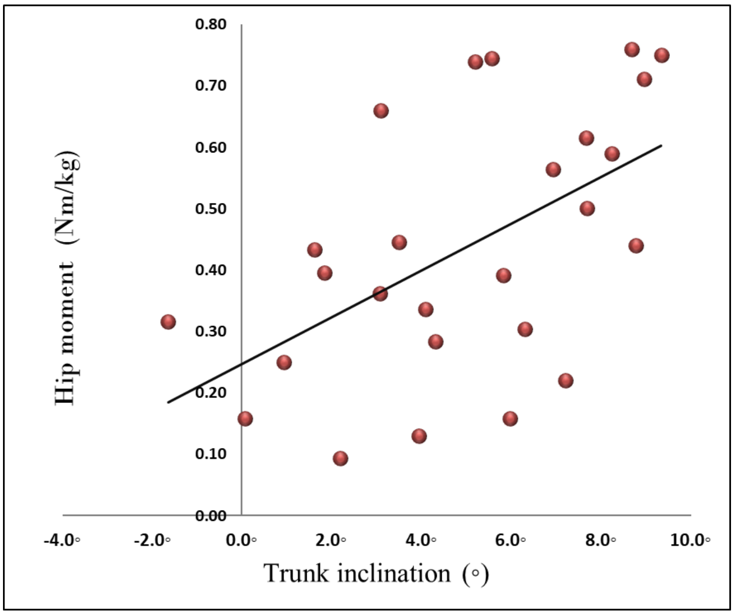


Figure 5-2. Correlation between trunk inclination and hip moments averaged across a specific period of stance phase (15-25%) during walking for knee OA group.

The plots above show that there was considerable variability across the two samples (knee OA and healthy). Although there was a trend for hip moment to increase as trunk inclination increased, there were some cases in which higher trunk inclination was associated with relatively low hip moments. Nevertheless, in general the subjects with higher trunk inclination appeared to walk with increased hip moments. This idea is highlighted in the plots below which show example participants with high/low trunk inclination and the corresponding hip moment curves. These examples have been shown both for two healthy subjects (see Figure 5-3) and also two subjects with knee OA (Figure 5-4).

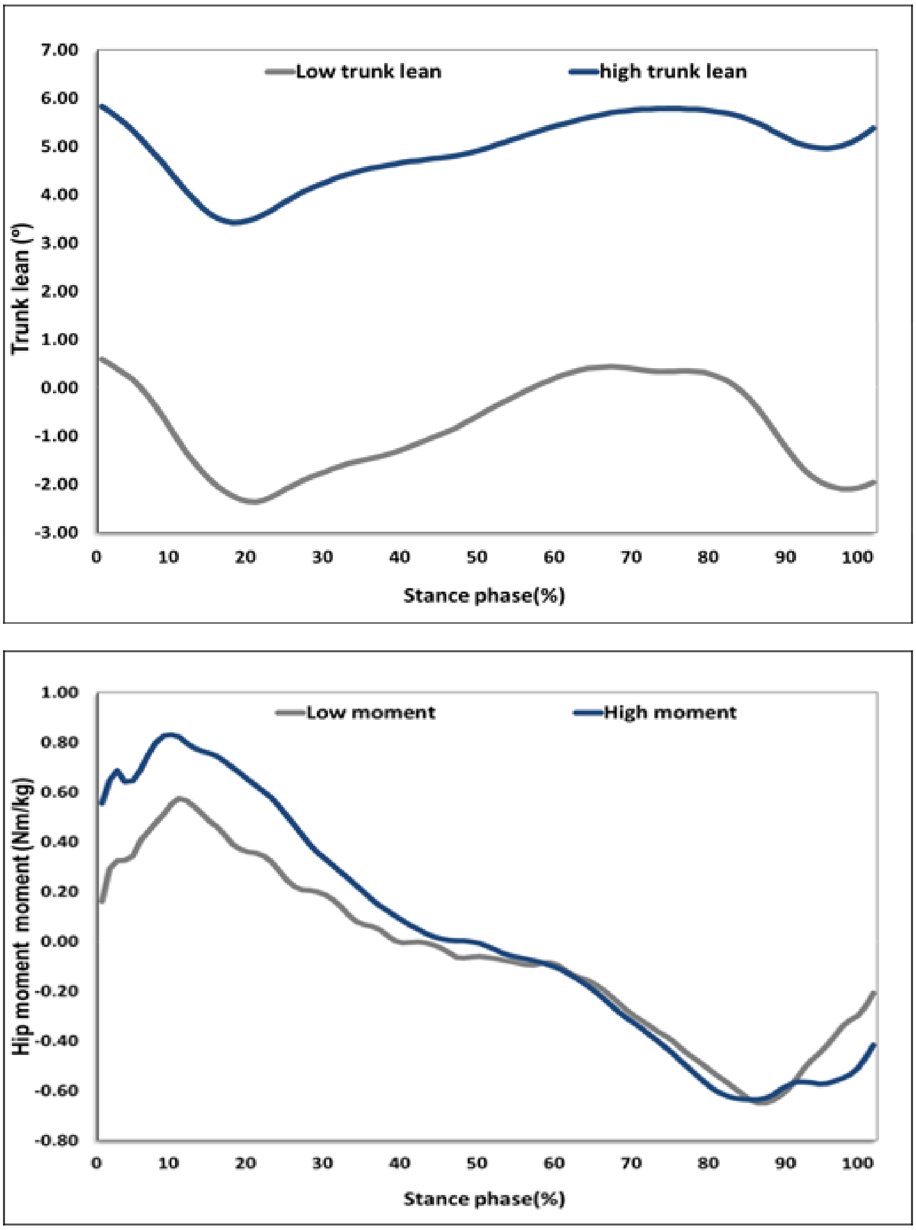


Figure 5-3. Example plots of variability of trunk inclination and hip moment (high trunk lean/high hip moment & low trunk lean/low hip moment) for healthy group.

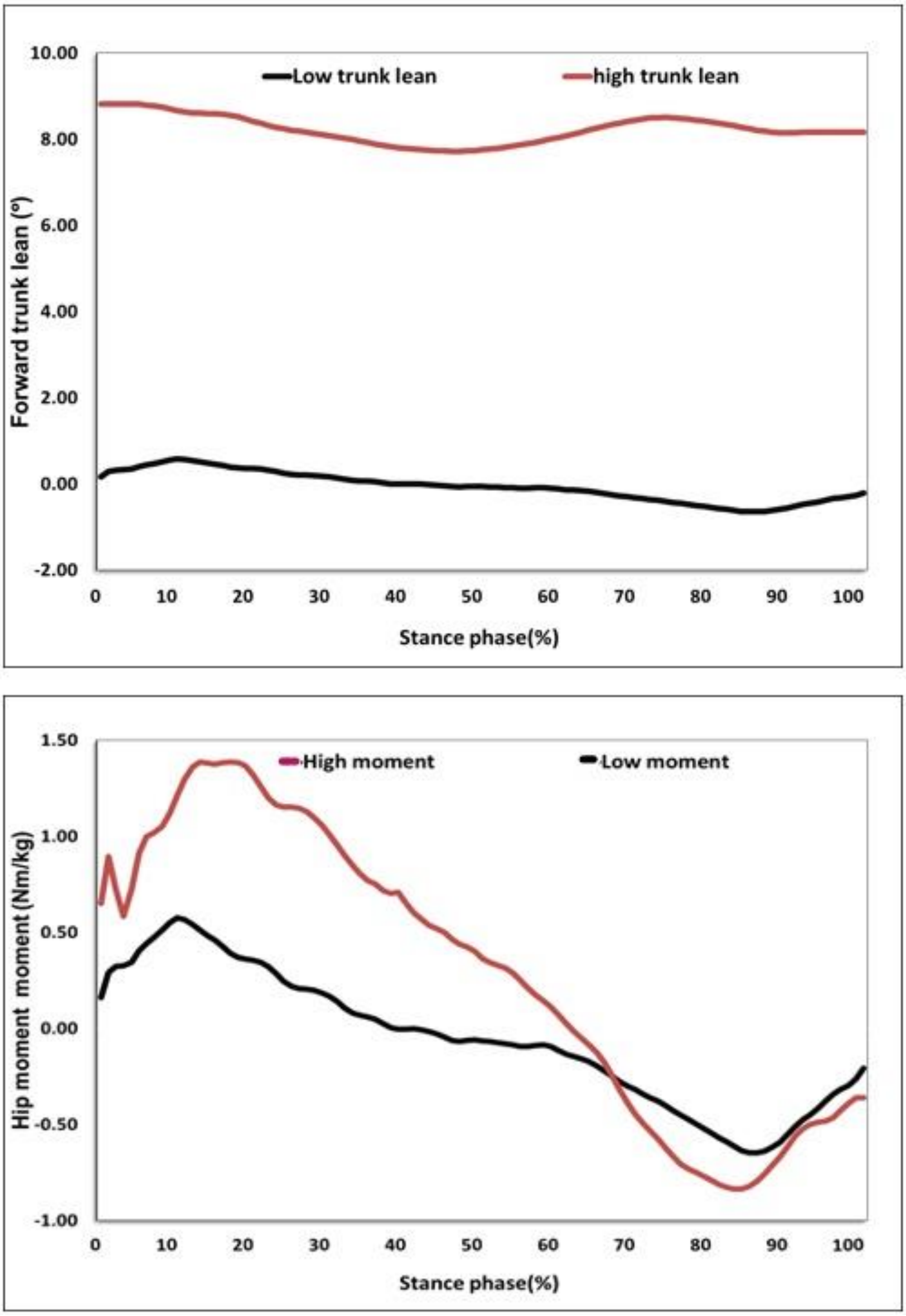


Figure 5-4. Example plots of variability of trunk inclination and hip moment (high trunk lean/high hip moment & low trunk lean/low hip moment) for knee OA group.

5.4.2 Relationship between trunk inclination and hamstring /quadriceps /gastrocnemius activity

Relationship between trunk inclination and hamstring muscle activity

Based on the correlation guidelines in Table 5-1, the results showed that there was a positive weak correlation between trunk inclination and the combined biceps femoris and semitendinosus muscle (hamstrings) in the healthy group (Table 5-3). Note that this correlation was performed for the combined muscle activity averaged over the period 15-25% of stance.

Surprisingly, no correlations were found when the muscle values were tested individually, nor was there a correlation in the group with knee osteoarthritis. Inspection of the individual correlation plots showed that this lack of a statistical correlation was the result of wide variability across the cohort.

Table 5-3. Results for association between forward trunk inclination and hamstring (medial and lateral) activity for healthy, knee OA and combined group from 15-25% of stance phase.

Variable	**Group**	**Control shoes**	
		R	***P***
Relationship between trunk inclination and biceps activity	***Healthy***	.28	.25
	OA	-.36	.86
	Combined	.17	.25
Relationship between trunk inclination and semitendinosus activity	***Healthy***	.10	.70
	OA	-.17	.38
	Combined	.19	.18
Relationship between trunk inclination and combined hamstring muscle activity (biceps and semitendinosus muscles)	***Healthy***	.47*	.04*
	OA	-.12	.52
	Combined	.200	.183

Relationship between trunk inclination and quadriceps muscle activity

The combined vastus medialis and vastus lateralis values showed a positive moderate association with forward trunk inclination for the healthy group (r=0.47), but not for the group with knee OA. However, again, no correlations were seen with the individual quadriceps muscles. The statistical results for this section are presented in Table 5-4.

Table 5-4. Results of association between forward trunk inclination and quadriceps muscle activity (VM, VL and combined muscles) for healthy, knee OA and combined groups from 15-25% of stance phase.

Variable	Group	Control shoes	
		R	**P**
Relationship between trunk inclination and VM activity	***Healthy***	**.162**	**.52**
	OA	**-.19**	**.33**
	Combined	**-.06**	**.68**
Relationship between trunk inclination and VL activity	***Healthy***	**.36**	**.14**
	OA	**-.064**	**.75**
	Combined	**-.12**	**.14**
Relationship between trunk inclination and combined muscle activity (VM & VL muscles)	***Healthy***	**.47***	**.04**
	OA	**-.12**	**.52**
	Combined	**-.017**	**.911**

Relationship between trunk inclination and gastrocnemius muscles

This correlation analysis showed that there was no relationship between activity of the gastrocnemius muscles and trunk inclination in walking across any of the groups. Table 5-5 shows the results of this analysis.

Table 5-5. Results of association between forward trunk inclination and gastrocnemius activity (medial, lateral and combined) for healthy, knee OA and combined groups at 15-25% of stance phase.

Variable	Group	Control shoes	
		R	P
Relationship between trunk inclination and medial gastrocnemius activity	***Healthy***	**-.07**	**.76**
	OA	**-.04**	**.81**
	Combined	**.07**	**.63**
Relationship between trunk inclination and lateral gastrocnemius activity	***Healthy***	**-.12**	**.63**
	OA	**.10**	**.59**
	Combined	**.15**	**.31**
Relationship between trunk inclination and combined gastrocnemius muscles activity (medial & lateral gastrocnemius)	***Healthy***	**-.065**	**.793**
	OA	**.054**	**.788**
	Combined	**.154**	**.308**

5.4.3 Relationship between trunk inclination and muscle co-contraction

Mean values for co-contraction of four pairs of muscles around the knee joint were investigated for correlation with trunk inclination in walking. However, consistent with the findings presented above of minimal correlations between muscle activation levels and trunk inclination, no meaningful correlations were observed between any of the co-contraction measures and trunk inclination across the period 15-25% of stance phase (Table 5-6).

Table 5-6. Summary of the correlation results between forward trunk inclination and co-contraction.

Variable	Group	Control shoe	
		R	***p*-value**
Relationship between trunk inclination and BF-VL co-contraction	Healthy	.23	.36
	OA	-.06	.74
Relationship between trunk inclination and semitendinosus -VM co-contraction	Healthy	.29	.24
	OA	-.22	.29
Relationship between trunk inclination and lateral gastrocnemius - VL co-contraction	Healthy	.31	.19
	OA	-.32	.08
Relationship between trunk inclination and medial gastrocnemius -VM co-contraction	Healthy	.13	.60
	OA	-.18	.34

Relationship between trunk inclination and combined co-contraction

Correlational analysis was performed on the combined muscle signals (obtained by summing the medial and lateral muscles) for each of the three muscle groups studied. Consistent with the data presented above, no meaningful correlations were observed between forward trunk inclination and hamstring-quadriceps co-contraction or between forward trunk inclination and gastrocnemius-quadriceps co-contraction for any of the groups (healthy, knee OA and combined groups). This data is presented in Table 5-7 below.

Table 5-7. Summary of the correlation results between forward trunk inclination and hamstring-quadriceps and gastrocnemius-quadriceps co-contraction.

Variable: Relationship between trunk inclination and:	Group	**Control shoes**	
		R	**P**
Hamstring-quadriceps co-contraction	***Healthy***	**.16**	**.50**
	OA	**-.15**	**.45**
	Combined	**.04**	**.76**
Gastrocnemius-quadriceps co-contraction	***Healthy***	**-.02**	**.92**
	OA	**-.11**	**.57**
	Combined	**.02**	**.88**

5.5 Discussion

5.5.1 Overview

This study aimed to investigate correlations between sagittal plane trunk inclination, sagittal moments, muscle activity and co-contraction while walking. In order to do this, I focused on a specific period of the gait cycle (15-25% of stance) as this has been shown to correspond to the period of peak loading. The motivation for this study was based on the ideas presented in the literature review on how trunk inclination could affect joint moment and how, in turn, this may affect muscle activations. The study follows from the previous chapter in which I observed a significant increase in forward lean in the participants with knee osteoarthritis. The findings of this study showed a weak to moderate association between forward trunk inclination and sagittal hip moment. The data also showed a weak correlation between forward trunk inclination and combined hamstring activity along with a weak correlation between combined quadriceps activity and trunk inclination. However, these links were only observed for the healthy subjects and were not evident in the data on the individual muscles: nor was there any link between co-contraction and trunk inclination.

There has been minimal previous work which has investigated links between trunk lean and biomechanical characteristics of the lower limb. Therefore, direct comparison with other studies is not always possible. However, in this section, the research questions and their related findings will be systematically discussed and linked to previous studies wherever possible. The limitations of the study will then be discussed, followed by conclusions, the overall implications of the study and recommendations for further research.

5.5.2 Correlation between forward trunk inclination and lower limb moments

The findings for the association between forward trunk inclination and sagittal hip moment showed a moderate positive correlation in the both the knee OA and healthy groups, while a weak correlation was found for the combined groups. Thus, although not a strong correlation, this does suggest that trunk inclination impacts upon hip moments, as predicted in a modelling study by Kagaya et al. (2003) and suggested by Sato and Maitland (2008) and Leteneur et al. (2009) when studying healthy subjects.

Small changes in the inclination of the segment have the potential to influence the moment arm of the GRF vector around the hip joint centre. Specifically, as the centre of mass (CoM) is shifted forwards, in relation to the hip centre, the external hip extensor moment will increase. This external moment will require an increased internal moment from the action of the hip extensor muscles to maintain the upright position of the trunk segment during walking. Although the findings of study two support this hypothesis, the issue remains as to why the correlations found were relatively weak.

Based on the mechanism outlined above, I hypothesised that the data would show a strong link between greater forward trunk lean and increased sagittal hip moment. Equally, I anticipated that individuals with lower forward inclination of the trunk would consistently show lower hip moments, as seen in the example plots given in results section of this chapter (Figures 5-3 and 5-4). However, only moderate correlations were observed. The various factors which may account for this inconsistency are explained below. However, before these factors are discussed in detail, it is important to first reflect on the reliability of the data presented in the methods chapter of this thesis. These data showed that all kinematic data were robust to potential errors which could have resulted from inconsistencies in marker placement. I am therefore confident that the weaker than expected correlation was not the result of errors in the experimental data.

Nevertheless, precisely quantifying the position and motion of the trunk can be problematic and the issues associated with this type of measurement are discussed below.

Unlike the shank and thigh segments, the trunk is not a rigid body segment, but a highly complex structure with multiple articulations. In order to produce a useable kinematic model of the trunk (and therefore an estimate of trunk inclination), it was necessary to make choices about how to create a single rigid-body model of this segment. I selected a model of the trunk in line with the recommendations of Armand et al. (2014). With this model, the trunk segment was defined using markers on the left/right acromiums and left/right greater trochanters and tracked using markers placed on the IJ and on T2 and T8.

Small differences in bony anatomy and/or relaxed standing position have the potential to create an offset in sagittal trunk inclination. Difficulties emerge because individual people have varied anatomical characteristics: for instance, they may show more or less rounding at the shoulder, unbalanced muscles in the upper part of the trunk or weakness in the spinal muscle or trapezius, which could lead to subtle, but important, differences in anatomical alignment. Such differences between people will affect the definition of the anatomical coordinate frame and could lead to an offset in sagittal trunk inclination which would cause the trunk inclination data to be shifted either upwards (increased trunk inclination) or downwards (decreased trunk inclination). Given the relatively small range of trunk inclination across the cohort, small changes in individual trunk inclination, due to subtle differences in anatomical alignment, will increase the spread of the data in the correlation analyses. This source of variability could have led to the weaker than anticipated correlations.

Uncertainty in measuring hip moment

A further biomechanical factor for which there could be measurement uncertainty is hip moment. Although my repeatability study demonstrated good reliability for the hip moments, this only indicates that the measurement was repeatable. It does not preclude the possibility that there might have been some aspects of the measurement which might have resulted in a consistent error. The knee OA group, and matched controls, in this study had a higher body mass index than would be expected in a younger group. This increased body mass index is the result of excess adipose tissue which is associated with increased soft tissue movement artefacts (Cappozzo et al., 1996), difficulty in locating the precise position of the hip joint centre (Kainz et al., 2015) and tracking the pelvic segment (Becker and Russ, 2015). Errors in the anatomical definition and kinematic tracking of the pelvis and thigh segment will translate into errors in the calculation of the hip joint moment which is derived using the segmental motions and the precise position of the joint centre relative to the direction of the ground reaction force vector. As explained above, uncertainty in one of the variables of the correlational analysis (hip moment) will lead to reduced correlation due to more random noise and therefore more scatter of the data. Thus, soft tissue artefacts and the associated random error in hip moments may have contributed to the lower than expected correlation between trunk inclination and hip extensor moment.

Turning to the findings for knee and ankle moments, no correlations were observed between these moments and the degree of forward trunk inclination. This could be because moment is determined by the magnitude of the GRF vector and the perpendicular distance between this vector and the knee/ankle joint centre. While small changes in trunk position, may lead to relatively large changes in the distance between the GRF vector and the hip joint centre, it is possible that such alterations in trunk position may not have a major effect on the distance between the GRF vector and knee/ankle joint centre. This may explain that lack of a correlation between trunk inclination and knee/ankle moments. Alternatively, the sources of variability in the measurement, described above, may have introduced noise into the measurements, which would weaken any possible correlation.

A number of previous studies have investigated the effect of altering upper body position on lower limb moments in healthy subjects. However, most of these studies have instructed subjects to adopt an unrealistic trunk inclination, not characteristic of normal walking and therefore it is not possible, in most cases, to make meaningful comparisons of my results with this work. Nevertheless, it is important to reflect on the findings of these studies in the context of my results. One recent study (Kluger et al., 2014) showed that knee and hip moments joint moments change significantly when trunk flexions of 25 ±7° and 50±7° are adopted,. They also found that ankle plantarflexion moment was decreased with increasing trunk flexion.

Lewis and Sahrmann's (2015) study examined trunk flexion effects on kinematic and kinetic lower limb measures. They instructed healthy participants to walk using first, their natural posture, then with a swayback posture and finally while inclining the trunk forwards (with flexed knees and hips). The findings showed that walking with the swayback position led to increased hip flexor moment, while when walking with a forward flexed posture, hip extensor moment increased. This is consistent with the finding observed in this present study. Kluger et al. (2014) studied the impact of a sustained trunk inclination on kinetic variables in the lower limbs for healthy subjects while walking, in which participants walked first using an upright posture, then with trunk flexion of 25° ± 7, and then at 50° ± 7degree. It was found that as trunk flexion increased, plantar flexor moment was reduced, while peak hip extensor moments were increased. However, these trunk inclinations are very large and therefore this study is not directly comparable with the data presented in this thesis.

Leteneur et al. (2009) divided twenty-five healthy participants into two groups based on trunk inclination, with one group containing those who leaned backwards naturally (average inclination -1.7°) and the other those who leaned forward (average inclination 2.9°), and investigated lower limb moments. The results demonstrate clear differences in hip moments between the groups, with duration of hip extension moments being increased for the forward group. These patterns match those observed in the current study, however they did not report correlations and therefore direction comparison is not possible. Interestingly, differences in moments were less clear at the knee and ankle, which is again consistent with the findings presented in this thesis.

5.5.3 Correlation between forward trunk inclination and muscle activity

The next research question in the study investigated correlations between forward trunk inclination and activity of the muscles of the lower limb. The data did not show a significant correlation between trunk inclination and either biceps femoris or semitendinosus muscles when they were measured individually. However, when both hamstring muscles were combined, a weak correlation was observed with trunk inclination. in the healthy group but not the group with knee osteoarthritis. Thus, although these data provide some support for the idea that forward trunk lean will be associated with increased hip extensor muscle activity, the results were not conclusive and therefore it is important to consider the other factors which may have influenced this relationship.

Although most previous researches into hamstring patterns in people with knee osteoarthritis (Childs et al., 2004, Rutherford et al., 2017, Sharma et al., 2017) has focused on the hamstring muscles individually, I chose to combine them as part of the analysis. The findings that, together, there was a correlation appears biomechanically plausible as the hamstring will work as a group to create an extensor moment at the hip. Therefore, increased trunk lean could be compensated for by the two muscles acting together, rather than by one of the hamstring muscles providing the necessary increase in force. Nevertheless, the correlations were not strong in the healthy group and there were no meaningful correlations observed in the group with knee OA. This lack of a correlation could have resulted from the synergistic nature of the hamstrings and the gluteus maximus muscles, which also act to extend the hip. The analysis described above was limited to the hamstring muscles. However, it is possible that increases in trunk inclination were compensated for by increases in gluteus maximus activity which were not measured in this study. This methodological choice was made because of the difficulty in obtaining high fidelity signals from this muscle during walking in people with increased levels of adipose tissue. Therefore, further research is needed to understand the link between trunk inclination and gluteus maximus activity.

As explained in the section above, there are a number of limitations which result from modelling the trunk as a single segment. These limitations may lead to uncertainty in the measurement of trunk inclination and therefore increased scatter in the data. This scatter will reduce possible correlations and might explain why the correlations between the trunk inclination data and the EMG data were lower than anticipated. Furthermore, EMG is a complex measurement which is also associated with a level of uncertain, especially when comparisons are to be made across different subjects. Again, this uncertainty, or noise, could have lowered observed correlations.

The findings for the quadriceps revealed no correlation between VM muscle activity and trunk inclination in either of the groups. This finding was similar for the VL muscle also, with no correlations found. However, there was a moderate positive correlation found for the combined activity of the quadriceps and forward trunk lean in the healthy but not in the knee OA or combined groups. This finding of a correlation may be explained as a part of the mechanism of forward lean. It is possible that the increases in hamstring moment, as well as acting to increase upper body position, may also act to increase the flexor moment at the knee. A corresponding increase in quadriceps activity would act to balance this moment at the knee, resulting in no change in knee moment. Interestingly, the knee OA group in this study did not demonstrate a correlation between quadriceps activity and trunk inclination. However, there was no correlation between hamstring activity and trunk inclination, suggesting different mechanisms underlying neuromuscular control. No correlations were observed between trunk inclination and either medial or lateral gastrocnemius activity, and further, no correlation was seen with the combined gastrocnemius results in either healthy or knee OA group. However, given the lack of a correlation between ankle joint moment and trunk inclination, this result could be expected.

It is interesting to compare the muscle activity data in this study which that observed by Grasso et al. (2014), as the only previous study to investigate how muscle activity during walking may change with trunk lean. In this study, five healthy participants aged 21-36 were tested while walking at in three different postures; regular, with flexed knee, and with trunk and knee flexed. However, it is important to note that forward leans of up to 50º were adopted by the participants in their study and therefore direct comparison of the findings with this present study are difficult. Given the small number of people in the study by Grasso et al. (2014), it was not possible to make generalisable statements; however, their data did suggest a general increase in muscle activity as trunk inclination was increased. These findings are consistent the hamstring data reported in this current study over a more modest range of trunk inclination angles.

5.5.4 Correlation between forward trunk inclination and muscle co-contraction

The analysis revealed no correlations between trunk inclination and any of the co-contraction indices studied. However, given that there was only a weak link between hamstring activation and trunk lean, and minimal other correlations between trunk inclination and muscle activation, this lack of a correlation between co-contraction and trunk inclination was inevitable. Increased co-contraction activity between medial knee muscles has been linked by Hodges et al. (2015) to more rapid progression of knee OA. Therefore, as part of the background developed in the literature review section, I suggested a hypothesis around a possible link between trunk inclination and co-contraction. The key idea was that increased trunk inclination may be associated with increased co-contraction and therefore increased compressive force at the joint.

The lack of a clear correlation between co-contraction and trunk inclination may indicate that joint contact forces are not influenced by upper body position. However, it is also possible that measurement uncertainty in both trunk inclination (see section 5.5.2) and EMG measurement, increased the scatter in the data, thereby hiding potential correlations. As explain earlier, the range of trunk inclination across the cohort was relatively small (1.6 º ±3º for healthy and 4.6º ±2.9º) for knee OA and so relatively small errors in this parameter could have led to relatively high levels of scatter which would have hidden a true correlation. EMG data is also problematic, especially as this needs to be normalised to an MVIC. Given that co-contraction is calculated from two (potentially noisy) EMG variables, this could have led to an increase in the scatter of the data, again hiding a potential link between trunk inclination and co-contraction.

5.5.5 Clinical implications of the findings

The most important findings from the study are the weak-moderate correlations between trunk inclination and hip moment and the moderate correlation observed between trunk inclination and hamstring activation. While these findings were not strong, they do provide some support to the idea that upper body position may influence hip moments and potentially hip extensor muscle patterns. If such findings are strengthened by further studies, it is possible that clinical strategies for reducing forward inclination for people with knee OA may be a viable treatment option for reducing muscle activity and therefore improving pain and function. Such clinical approaches may include stretching exercises for the hip flexor muscles, postural re-education and specially designed footwear. However, further intervention studies are required to establish if such approaches could improve trunk position and if this could lead to a corresponding change in muscle activation.

5.5.6 Limitations of the study

As discussed in Section 5.5.2, it is challenging to quantify forward trunk inclination, due to issues in kinematic modelling this a multi-articular structure, which is not a rigid segment. To ensure that data collection methodology was sound, I performed a repeatability study which showed high levels of repeatability for the measurement of the thoracic segment (see chapter on methods). However, this only indicates that the measurement was repeatable. It does not preclude the possibility that there might have been some aspects of the measurement which might have resulted in a consistent error, such as differences in anatomical alignment in the standing position. Such systematic errors could not have been eliminated completely and may have affected the findings of the study. As explained earlier, these uncertainties could have had a large influence on the derived correlations.

5.5.7 Conclusions

This second study aimed to establish whether joint moments, muscle activity and co-contraction were correlated with increased forward trunk inclination. While the findings did not show the strong correlations, this does not necessarily imply that there is no link between upper body position and muscle activations/co-contraction. Indeed, I suggest that for some variables, particularly trunk inclination, there were inherent difficulties in measurement which are difficult to avoid but which may have hidden a true correlation. Nevertheless, the findings offer some interesting implications, showing a weak-moderate correlation between trunk inclination and hip moment for both groups and a weak correlation between trunk inclination and hamstring activation in the healthy group. These findings show the beginnings of a link between upper body position and joint loading, but further research is required to develop more insight into this phenomenon.

Chapter 6 - Study three: The biomechanical effects of rocker footwear in people with knee OA

Overarching aim:

To investigate the effect of a footwear intervention (the three-curved rocker shoe) on biomechanical variables related to joint loading, both in people with knee OA and healthy control subjects.

6.1 Overview of the study

This is the final study of this Ph.D. thesis. This study aims to investigate the biomechanical effect of the three-curve rocker shoe on trunk inclination, lower limb moments and muscle activation patterns during gait with a focus on the early stance period, from 15-25% of stance. While previous studies of knee OA groups have tested lateral wedges and other shoe designs, most studies have focused on the reduction in knee adduction moments as a desired outcome. This study is novel in focusing instead on a footwear intervention to reduce moments in the sagittal plane, muscle co-contraction around the knee and trunk inclination. In the literature chapter, a rationale was put forward for how greater forward trunk inclination might have an impact on both muscle co-contraction and moments in the sagittal plane. In the previous chapter, the observed correlations between trunk inclination and other biomechanical variables was shown to be weak. However, I argued that this weaker-than-expected correlation may have be a result of uncertainty in measurement of trunk inclination which could have resulted from anatomical variation. Therefore, in order to explore further the potential link trunk inclination and lower limb biomechanics, it is interesting to use a within-subject design in which the effects of an intervention, which has the potential to change trunk inclination, are examined. For this study, a simple footwear intervention was chosen: the three-curve rocker shoe.

Rocker shoes have a curved, inflexible outsole, which means that the foot rocks forwards in the stance phase of gait. Previous research has shown that rocker footwear can be used to alter trunk inclination (Ochsmann et al., 2016) and also that this approach may be an effective way to reduce hip moments (Buchecker et al., 2013). Significantly, a study by Buchecker et al. (2013) demonstrates the possibility that hip moment reductions for the early stance phase are achievable with a correctly designed rocker shoe. Based on the rationale proposed in this thesis, in which greater hip moment is linked with forward trunk inclination and increased activity in the hamstrings, a well-designed rocker shoe intervention might lead to reductions in values for these variables for a knee OA cohort.

The rocker shoe's effect on muscle activity is not clear, with some studies finding raised gastrocnemius activity (Forghany et al., 2014) and others showing decreases (Sobhani et al., 2013), and most research seeing little change in muscle activation (e.g. Santo et al., 2012). However, the effect of the exact shoe design should be taken into account here, and the appropriate outsole design for the rocker shoe may impact upon hamstring activity, and requires investigation. This study will investigate the effects of the rocker shoe on co-contraction in particular.

Considering trunk inclination, very few studies have examined the impact of rocker footwear on this variable. From these studies, Ochsmann et al. (2016) found that trunk inclination was significantly decreased by the rocker shoe tested, and this will be explored here, as these results raise the possibility that a rocker shoe design could lead to reductions in forward trunk inclination for people with knee OA. However, it is likely that the effect of the rocker shoe will be dependent on the precise geometry of the outsole.

Potentially, a very specific design of rocker shoe may be required to reduce trunk inclination, hip moments and co-contraction of the knee muscles in people with knee OA. Recently, Hutchins et al. (2012) proposed a rocker profile which contained three different curves, based on circles drawn around the ankle, knee and hip, and aiming to reduce the amount of muscle activity at the hip, while maintaining a normal walking pattern. This suggests that a design of this type could be suitable for testing with a knee OA cohort. The 3 curves designed into the shoe are positioned with the radius of each one corresponding to the sagittal plane centre of either the hip, knee or ankle joint during the stance phase of gait, allowing a gentle forward rolling motion of the shoe, intended by Hitchins et al. (2012) to reduce muscle activity for muscles of the knee and ankle around the knee and ankle. If hip extensor moment can be reduced also, it is possible that hamstring activity may reduce in line with this, reducing hamstring-quadriceps co-contraction.

The three-curve rocker profile used for this study has been designed to reduce both the hip and the ankle moment in early-mid stance and therefore may be an effective approach for decreasing both moments and also muscle activation and therefore co-contraction during 15-25% of stance phase. To date, this specific design of three-curve rocker shoe has not been tested in individuals with knee OA. However, there is a substantial amount of previous research demonstrating increased muscle activity/co-contraction during this period in people with knee OA. Therefore, if the three-curve rocker shoe is effective at reducing muscle activity, then it may prove an effective approach for the clinical management of knee OA.

This study will investigate the effect of three-curve footwear interventions (rocker shoe) on sagittal plane hip, knee and ankle moments and also muscle co-contraction patterns (hamstrings, quadriceps and gastrocnemius) in a cohort of patients with knee OA during walking during the early stance period (15-25%). The peak of compressive knee joint loading is identified by modelling studies as occurring at between 15% and 25% and 65-75% of the stance phase (Sritharan et al., 2016; Brandon et al., 2014). Both clinical and biomechanical outcomes will be collected as part of the study and a healthy control group included to understand whether the effect of the footwear differs between healthy people and those with knee OA.

6.2 Research Questions

RQ 3A: How does inclination change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?

RQ 3B: How do lower limb moments change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?

RQ 3C: How does muscular co-contraction change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?

RQ 3D: Are there immediate changes in measure of pain when people with knee OA wear a three-curve rocker shoe?

6.3 Methodology:

The methodology chapter should be referred to for a detailed description of the methods followed in each study for the thesis. This study was conducted with subjects wearing control shoes and then the intervention shoe (3-curve rocker shoe). All trials took place in the University's Gait Laboratory.

6.3.1 Sample and population

The sample for the third study of this thesis included 27 knee osteoarthritis patients and 20 healthy subjects, across both genders. The data for all three studies was collected in a single visit for both groups. The methods chapter provides full details of the criteria used for inclusion or exclusion of the subjects.

6.3.2 Design

A between-within subjects design (two-way mixed design ANOVA) was used for the present study. Therefore, this study was designed to examine the effect of two independent variables between and within groups. The first variable was the between-subject factor, which had two levels (healthy group and OA group). The second was the within-subject factor, which also had two levels (control and rocker shoes). In addition, this study included fourteen tested dependent variables within the subject design, as detailed in the outcomes section below.

6.3.3 Derivation of Outcome Measures and Statistical Methods

This study was designed to determine the association between use of a 3-curve rocker shoe and forward trunk inclination, lower limb joint moments and muscle co-contraction during the 15-25%. Outcome measures and statistical methods for the research questions set are presented in this section.

RQ 3a: How does inclination change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?

To answer this question, trunk inclination was recorded for all subjects when walking in the control shoes as a baseline and then when wearing the three-curve rocker shoe, as explained in the methods chapter and Chapter 4, Section 4.3.2. An average for the degree of forward inclination over 15-25% of stance phase was then calculated, to give a single value for each subject. The association between the values produced, the subject group and the footwear worn was then calculated two-way mixed design ANOVA.

RQ 3b: How do lower limb moments change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?

To address this question, mean sagittal moments for the ankle, knee and hip joints were calculated over the 15-25% period of stance phase. Two-way ANOVA was used to calculate the correlation between these values, subject type (knee OA or healthy group) and shoe type (control or rocker shoe).

RQ 3c: How does muscular co-contraction change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?

In order to examine the relationship between group, shoe type and lower limb co-contraction for the period 15-25% of stance phase, values for co-contraction were calculated for all subjects in both types of footwear, following the method described in Chapter 4, Section 4.3.2. The effect of group and shoe type, as well as interaction effects, on the different co-contraction values, were then calculated using two-way ANOVA analysis. I chose to use the sum method to quantify co-contraction as this has been shown to correlate better with joint contact loading (Sritharan et al., 2016; Hodges et al., 2016) than other approaches to calculating co-contraction (Heiden et al., 2009b).

RQ 3d: Are there immediate changes in measure of pain when people with knee OA wear a three-curve rocker shoe?

The Visual Analog Scale for Pain (VAS Pain) is a continuous scale, which is 10 centimetres in length (Huskisson, 1974, Hawker et al., 2011). The scale is used to measure pain intensity, with a score of 0 representing "no pain" and a score of 10 the "worst imaginable pain" (Jensen et al., 1986, Burckhardt and Jones, 2003, Ferraz et al., 1990). Specifically, from the VAS score, pain can be interpreted as severe at 7.5-10cm, moderate at 4.5-7.4 cm, mild at 0.5- 4.4 cm and not found at 0-0.4cm, based on prior studies of score distribution among post-operative patients (Jensen et al., 2003). The VAS pain scale was used to quantify the clinical outcomes of pain both before and after the footwear interventions.

6.3.4 Statistical Analysis

Statistical analysis of results was carried out using the statistical package for social sciences (SPSS) version 23 for Windows. The two-way mixed design ANOVA design investigated the individual effects of shoe type (control and rocker shoe) and subject (knee OA and healthy) for each dependent variable. ANOVA also allows these two independent variables to be compared against each other for interaction effects. This was used to investigate whether the effects seen from shoe type were affected by subject group (OA and healthy groups).

For between-subject analysis, homogeneity of variances and covariance were assessed by Levene's test and by Box's M test at ($p > .05$) respectively. For within subjects, Mauchly's test of sphericity was used. The assumption of sphericity was met for the two-way mixed design ANOVA ($p > .05$) for each dependent variable. All of these assumptions met the parametric analysis. Accordingly, two-way mixed design ANOVA was used to compare between subjects and within subjects. The LDS adjustment was used to investigate possible pairwise differences if the ANOVA showed a significant difference. Data are mean ± standard deviation, unless otherwise stated. The test was performed on the examined sample with the alpha level 0.05. Inspection of a box and whiskers boxplot of each of the tested variables prior to the analysis showed there were no outliers. The data was normally distributed, as assessed by Shapiro-Wilk's test of normality ($p > .05$). There was homogeneity of variances ($p > .05$) and covariances ($p > .05$), as assessed by Levene's test of homogeneity of variances and Box's M test, respectively. Mauchly's test of sphericity indicated that the assumption of sphericity was met for the two-way interaction. However, a non-parametric test was used to address RQ 3D. The VAS (pain score) as this is ordinal data.

The results section will present the main effects of the different shoe types, followed by the interaction observed. Although the ANOVA analysis provided statistics on the effect of group, this will not be discussed at length, as differences between knee OA and healthy subject groups were compared comprehensively in Chapters 4 and 5.

6.4 Results

6.4.1 Forward trunk inclination

When wearing the rocker shoe, the knee OA group showed an average reduction of 1.4° in forward trunk inclination compared to when wearing the control shoe. As seen from the ensemble curves in Figure 6-1, this difference remained consistent across the gait cycle. The ANOVA analysis for the mean trunk inclination value across 15-25% of stance showed main effects for footwear ($p=0.01$) and group ($p<0.01$) but no interaction (see Table 6). Specifically, there was a 30% reduction in forward lean in the OA group (effect size = .05) and a 31% reduction in forward lean in the healthy group. The finding of no interaction demonstrated that overall, the effects of footwear were similar for each group, albeit slightly smaller in the healthy group.

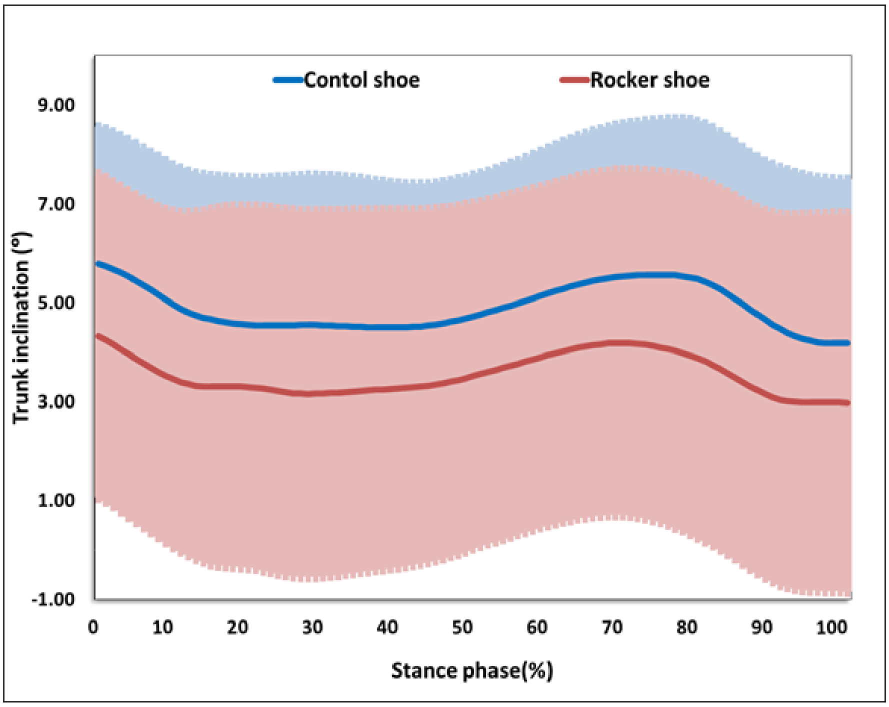


Figure 6-1. Ensemble average of trunk inclination for the knee OA group in two different shoes: (control shoes- blue line and (rocker shoes-red line) at stance phase.

Table 6-1. Summary result of trunk inclination before and after wearing rocker shoes in healthy and knee OA group.

Trunk inclination	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	P
OA group	4.6 ° (2.9)°	3.2 (2.7)°	0.5	• ***Effect of shoes=.014**** • ***Effect of group=0.005**** • ***Interaction effect=0.22***
Healthy group	1.6 ° (3)°	1.1 ° (2.7)°	0.17	

6.4.2 Lower limb moments

Sagittal hip moment

The ensemble curves for the different shoes showed only slight differences in sagittal hip moment (Figure 6-2). However, there was a consistent pattern across the stance phase for both groups. Specifically, for knee OA individuals, the curve showed a reduction in hip moment in the period between 20-70% of stance when wearing rocker shoes (Figure 6-2). Two-way mixed design ANOVA analysis for this factor across the period 15-25% stance phase found no significant main effects of shoes or group. No interaction effect was observed. Interestingly, mean hip moment was reduced by almost 8% in the knee OA group and almost 12% in the healthy group when wearing the rocker shoe (Table 6-2).

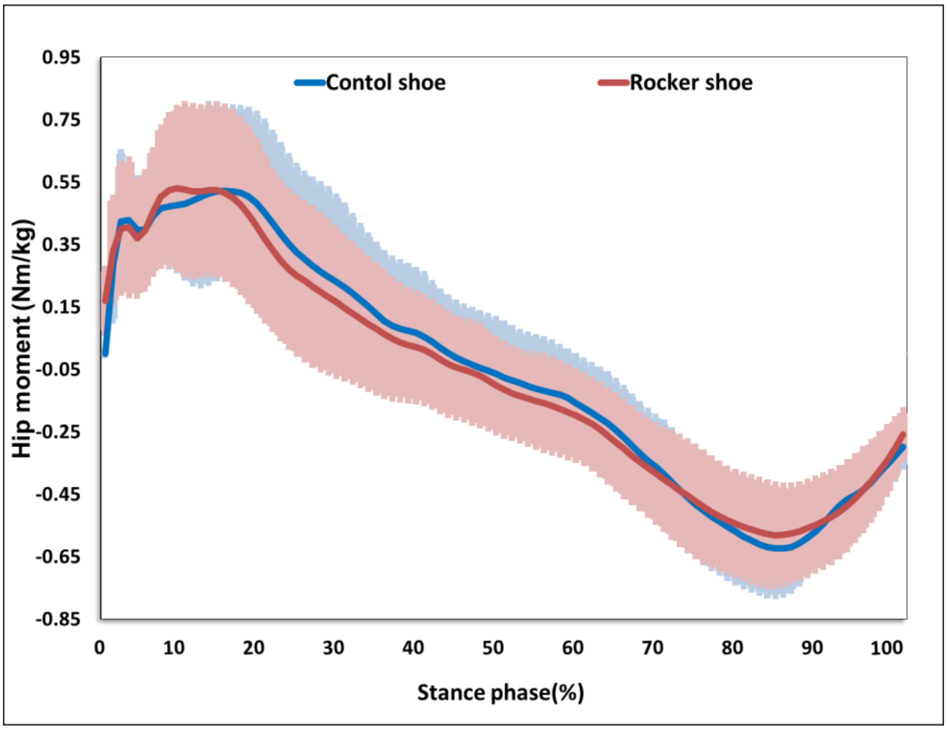


Figure 6-2. Ensemble average of sagittal hip moment for the knee OA group in two different shoes: (control shoes- blue line and (rocker shoes-red line) at stance phase.

Table 6-2. Summary result of sagittal hip moment before and after wearing rocker shoes in healthy and knee OA group.

Sagittal hip moment in:	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.39 (0.21)	0.36 (.24)	0.13	• ***Effect of shoes=0.11*** • ***Effect of group=0.27*** • ***Interaction effect=0.83***
Healthy group	0.34(0.20)	0.30 (.20)	0.2	

Sagittal knee moment

In the knee OA group, the ensemble curve for the rocker shoe demonstrated that sagittal knee moment was higher than for control shoes from 12-90% of stance phase, with a marked difference between 40 and 90% (Figure 6-3). No main effects were seen using ANOVA analysis across the period 15-25% of stance, however the effect of footwear approached significance (p=0.07). There were non-significant increases in knee moment, when both knee OA and healthy groups wore the rocker shoe, of 8% and 6% respectively. There was no interaction effect between footwear type and group (Table 6-3).

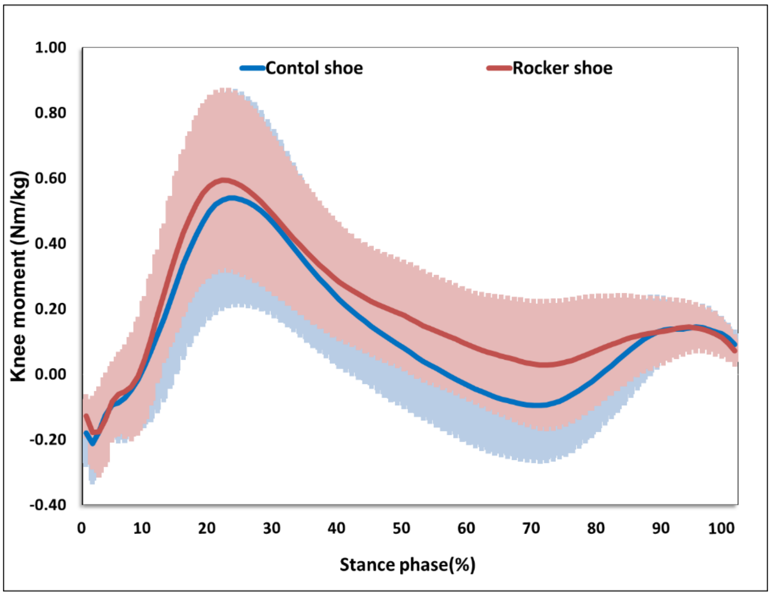


Figure 6-3. Ensemble average of sagittal knee moment for the knee OA group in two different shoes: (control shoes- blue line and (rocker shoes-red line) at stance phase.

Table 6-3. Summary result of sagittal knee moment before and after wearing rocker shoes in healthy and knee OA group.

Sagittal knee moment	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.51 (0.23)	0.55 (0.22)	0.18	• ***Effect of shoes=0.07*** • ***Effect of group=0.77*** • ***Interaction effect=0.89***
Healthy group	0.50 (0.18)	0.53 (0.18)	0.16	

Sagittal ankle moment

The two ensemble curves indicate that the rocker shoe reduced sagittal ankle moment at between 2-65% of the stance phase for the knee OA group, which included the period of interested, 15-25% stance (Figure 6-4). Two-way ANOVA results for mean ankle moment (15-25% of stance) demonstrated main effects for both shoes (0.012) and the group tested (0.032), without an interaction effect. Sagittal ankle moment decreased by 70% (effect size = 0.45) for the knee OA group and by almost 67% (effect size = 0.47) for the healthy sample. Lack of an interaction effect shows that the shoes had a similar effect for both healthy and knee OA subjects (see Table 6-4).

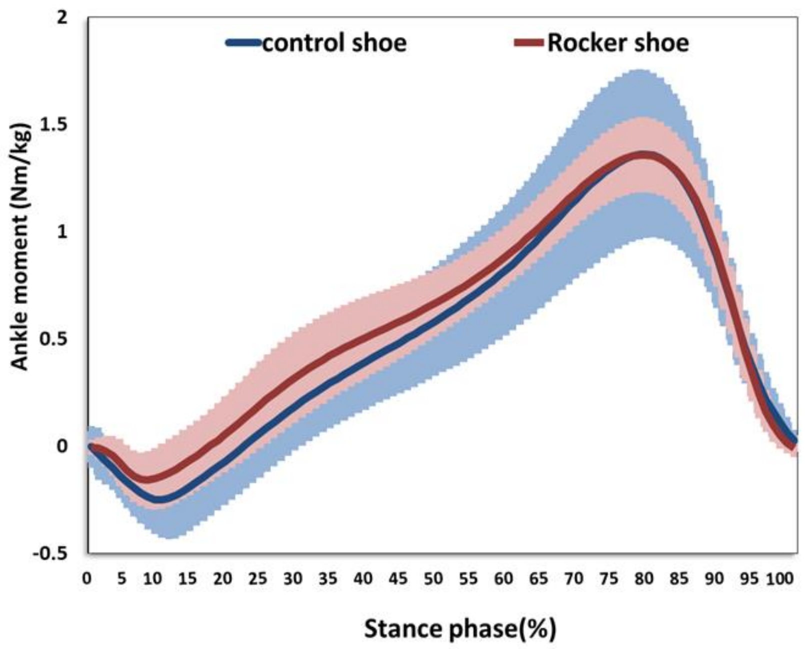


Figure 6-4. Ensemble average of sagittal ankle moment for the knee OA group in two different shoes: (control shoes- blue line and rocker shoes-red line) at stance phase.

Table 6-4. Summary result of sagittal ankle moment before and after wearing rocker shoes in healthy and knee OA group.

Sagittal ankle moment	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	-0.07(0.10)	-.021 (.12)	0.45	• ***Effect of shoes=0.012**** • ***Effect of group=0.032**** • ***Interaction effect=0.66***
Healthy group	-0.12 (0.10)	-.08 (.07)	0.47	

6.4.3 Muscle activity changes

Biceps femoris activity

The ensemble curve for biceps femoris activity in the rocker shoe shows clear differences with the control shoe for knee OA participants between 20 and 80% of stance phase, with a marked reduction between 30-60% of stance (Figure 6-5). When 2-way ANOVA was applied to the average BF activity values between 15-25% stance phase, no main effects of footwear were found (Table 6-5) and no interaction effect was identified.

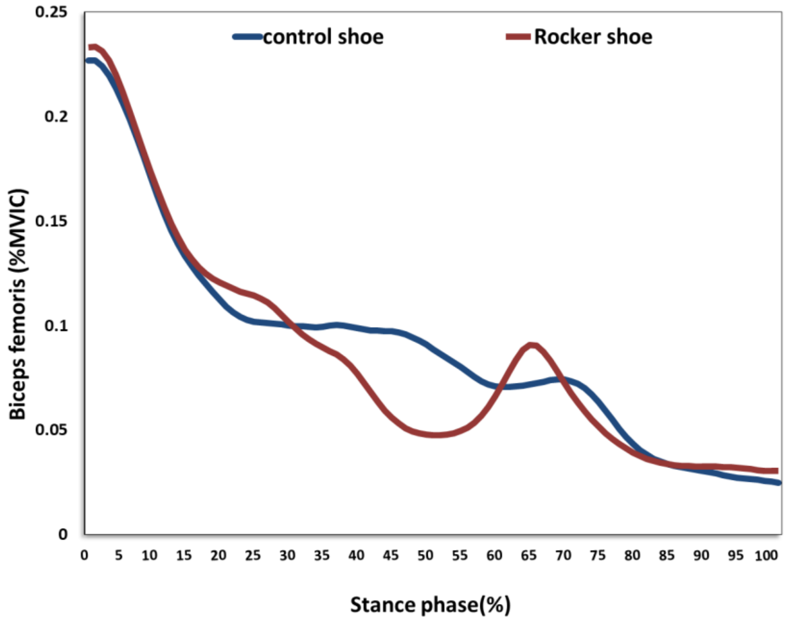


Figure 6-5. Ensemble average of biceps femoris muscle activity in stance phase for knee OA group within two shoes. (red-rocker, and blue –control shoes).

Table 6-5. Summary result of Biceps femoris activity before and after wearing rocker shoes in healthy and knee OA group.

Biceps femoris activity	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	*0.13(0.10)*	*.12 (.10)*	*0.1*	• ***Effect of shoes=0.51*** • ***Effect of group=0.10*** • ***Interaction effect=0.31***
Healthy group	*0.07(0.42)*	*.09 (.08)*	*-0.04*	

Semitendinosus activity

The ensemble curves reveal a consistent pattern, but with lower levels of semitendinosus activity for the treatment shoe from 0-85% of stance phase in the knee OA group (Figure 6-6). However, although the ANOVA analysis of mean values (across 15-25% stance) did not show a main effect for footwear (Table 6-6), there was an interaction, suggesting that medial hamstring activity was reduced, with the rocker footwear, in the group with knee OA but not in the healthy group.

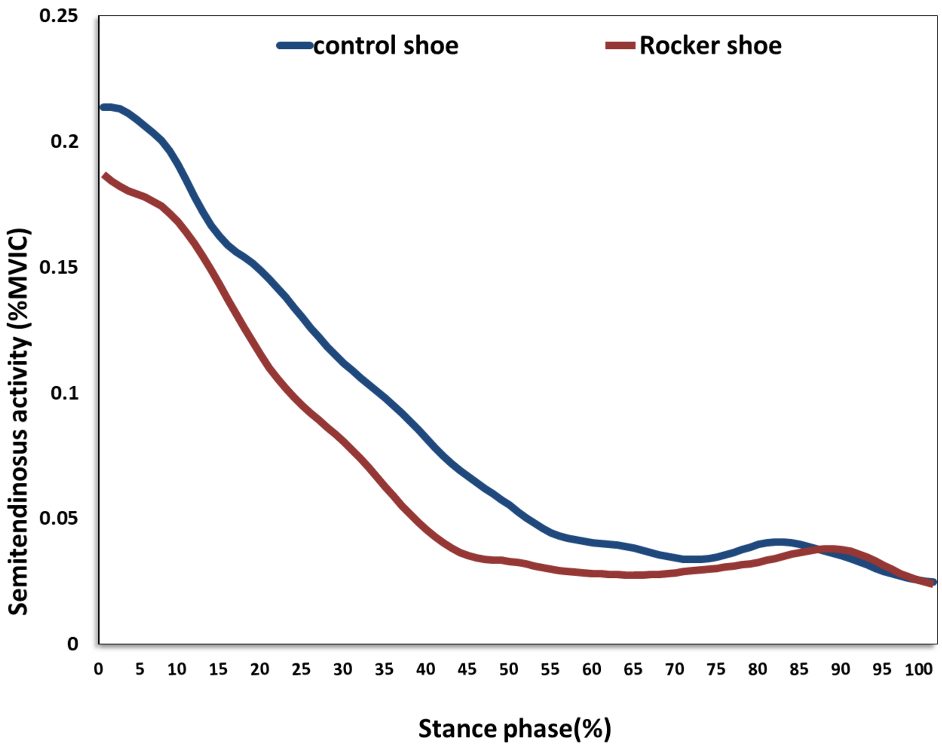


Figure 6-6. Ensemble average of semitendinosus muscle activity in stance phase for knee OA group within two shoes; (Red-rocker, and blue –control shoes).

Table 6-6. Summary result of Semitendinosus activity before and after wearing rocker shoes in healthy and knee OA group.

Semitendinosus activity	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.16 (0.12)	.13 (.12)	0.23	***Effect of shoes=0.22*** ***Effect of group=0.021**** ***Interaction effect=0.031****
Healthy group	*0.07(0.49)*	.08 (.06)	0.01	

Vastus medialis

Vastus medialis activity appeared to be higher between 35 and 70% of stance in people with knee OA, when they wore the with the rocker shoe compared to the control, (Figure 6-7). However, across the period of interest (15-25% stance), there was minimal change in the medial quadriceps activity with the different shoes and no evidence of any interaction effects (see Table 6-7).

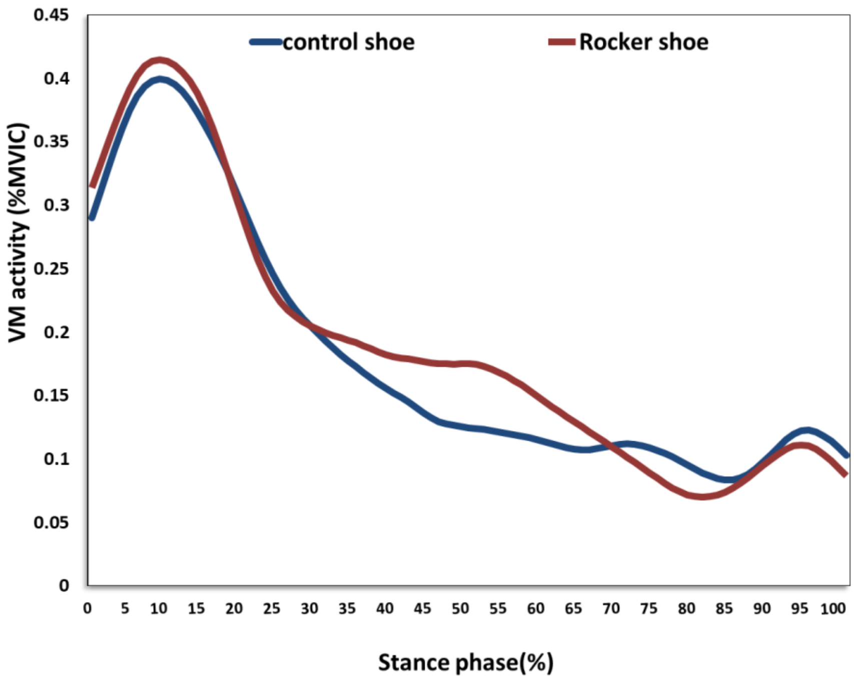


Figure 6-7. Ensemble average of VM muscle activity in stance phase for knee OA group within two shoes; (Red-rocker, and blue –control shoes).

Table 6-7. Summary result of VM activity before and after wearing rocker shoes in healthy and knee OA group.

VM muscle activity	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size (ES)	*P*
OA group	0 .36(0.21)	.37 (.33)	0.03	***Effect of shoes=0.66*** ***Effect of group=0.87***
Healthy group	0.34(.27)	.35 (.26)	0.03	***Interaction effect=0.99***

VL muscle activity

The ensemble curves shown in Figure 6-8 show slightly higher VL activity for the rocker footwear, in the people with knee OA, between 10 and 75% of stance phase. However, these differences were small, and the ANOVA analysis did not reveal any main effects of footwear across the period 15-25% of stance (Table 6-8), with an increase of only 6.5% for the knee OA group compared to 14% for the healthy group. No interaction effects were observed for this muscle.

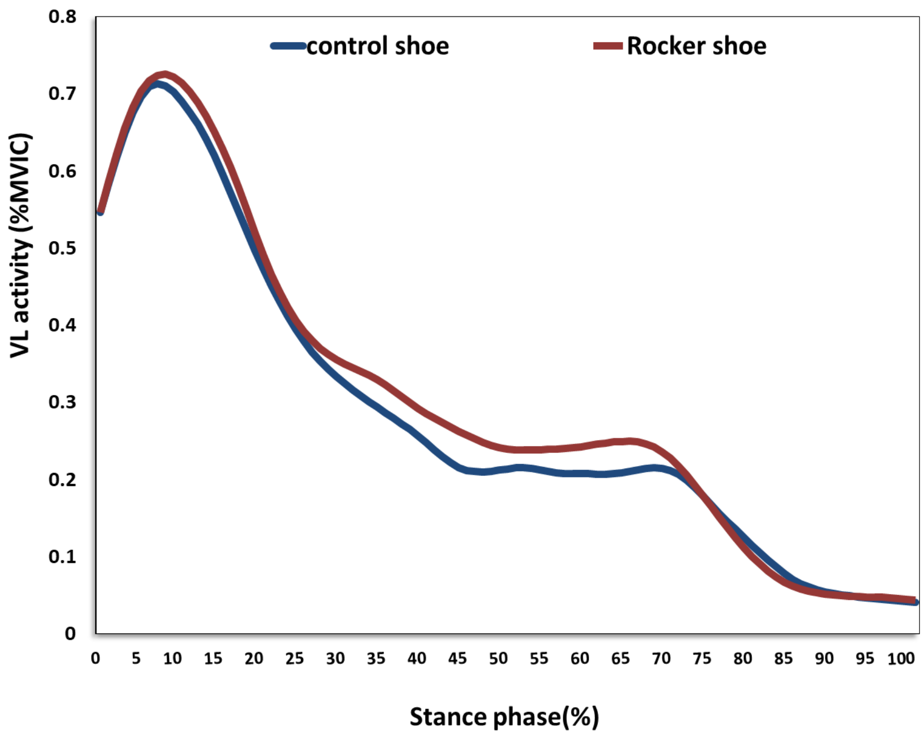


Figure 6-8. Ensemble average of VL muscle activity in stance phase for knee OA group within two shoes; (Red-rocker, and blue –control shoes).

Table 6-8. Summary result of VM activity before and after wearing rocker shoes in healthy and knee OA group.

VL muscle activity	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.61 (0.43)	.65 (.43)	0.09	• ***Effect of shoes=0.20*** • ***Effect of group=0.011**** • ***Interaction effect=0.89***
Healthy group	0.35 (0.31)	.40 (.18)	0.20	

Medial gastrocnemius activity

The knee OA group's ensemble curves for the two shoes tested show slight reductions in MG activity between 10 and 30% of stance and larger reductions between 30 and 75% (Figure 6-9). However, average values analysed by two-way ANOVA showed no main effects from footwear over 15-25% of stance phase (Table 6-9). These data indicate minimal change in MG activity for the healthy group compared with a 9% reduction for the knee OA group (effect size = 0.16). However, although these effects were slightly different, no interaction was observed between shoe type and subject group.

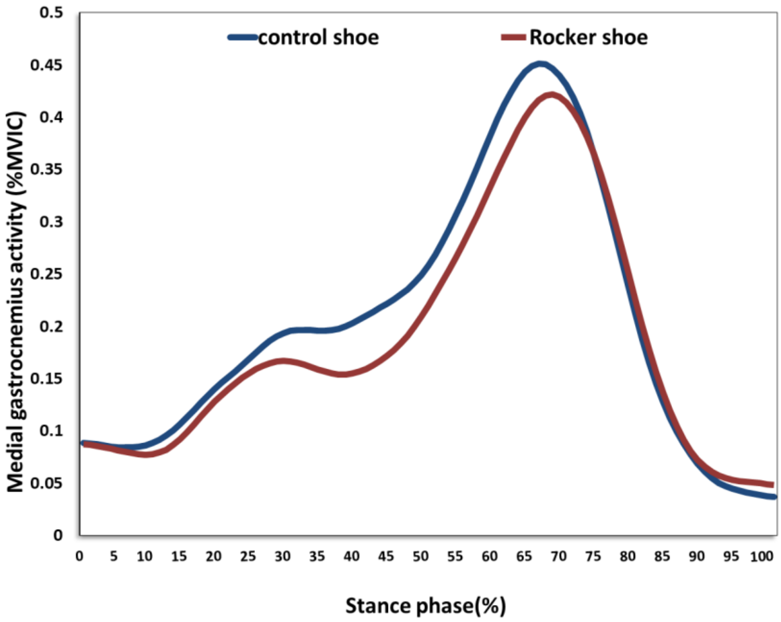


Figure 6-9. Ensemble average of medial gastrocnemius activity in stance phase for knee OA group in different shoes (red-rocker and blue –control shoes).

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Table 6-9. Summary result of MG activity before and after wearing rocker shoes in healthy and knee OA group.

Medial gastrocnemius activity	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.11 (0.06)	0.10 (.06)	0.16	• ***Effect of shoes=0.50*** • ***Effect of group=0.027**** • ***Interaction effect=0.83***
Healthy group	0.07 (0.39)	0.07 (.04)	0.00	

Lateral gastrocnemius activity

The ensemble curves for the two different footwear types show little difference in the knee OA group for LG activity (Figure 6-10). Given this similarity, the ANOVA analysis for average values across 15-25% of stance showed no effect of footwear and no interaction between the variables (Table 6-10).

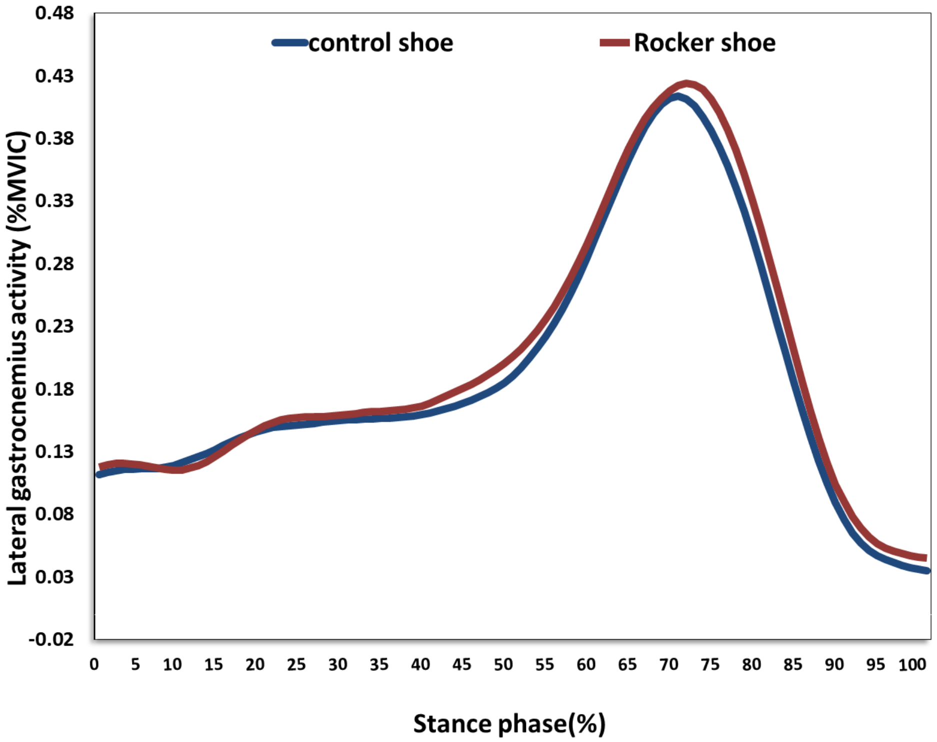

Figure 6-10. Ensemble average of lateral gastrocnemius activity in stance phase for knee OA group in different shoes (red-rocker and blue –control shoes).

Table 6-10. Summary result of LG activity before and after wearing rocker shoes in healthy and knee OA group.

LG activity	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.13 (0.12)	.13 (.10)	0.00	***Effect of shoes=0.12*** ***Effect of group=0.029**** ***Interaction effect=0.33***
Healthy group	.065 (.027)	.08 (.04)	0.09	

6.4.4 Muscle co-contraction

Biceps femoris-VL Co-contraction

The ensemble curves for the knee OA group shows a very similar trend for both footwear types, with the only clear differences between 58-75% of stance (Figure 6-11). Unsurprisingly, the ANOVA analysis, focused on 15-25% of stance, showed no significant effect of footwear (Table 6-11), although there was a clear difference between the two participant groups as discussed in Chapter 4.

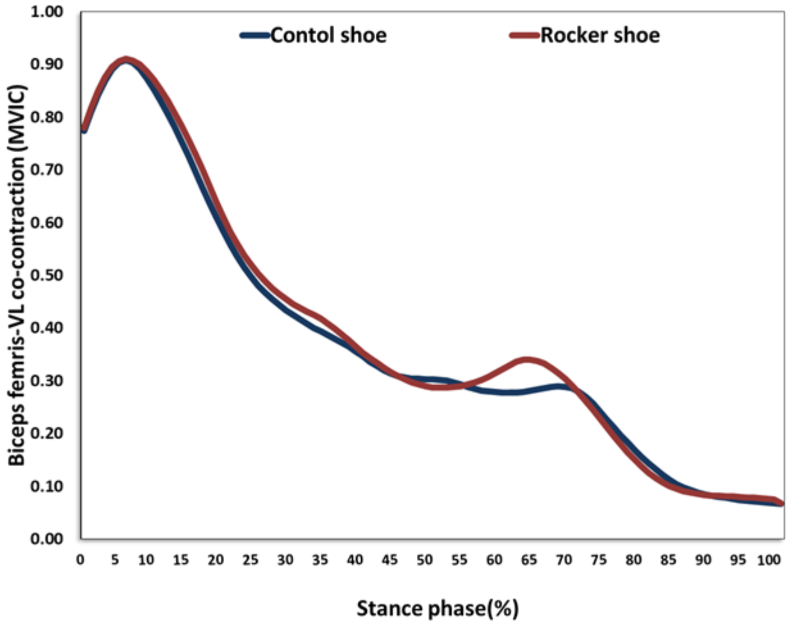

Figure 6-11. Ensemble average of biceps femoris and VL co contraction in stance phase for knee OA group in different shoes (Red-rocker and blue –control shoes).

Table 6-11. Summary result of biceps femoris-VL co-contraction before and after wearing rocker shoes in healthy and knee OA group (15-25%) stance.

BF-VL co contraction	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.74 (0.47)	0.77 (.47)	0.06	• ***Effect of shoes=0.86*** • ***Effect of group=0.001**** • ***Interaction effect=0.89***
Healthy group	0.41 (0.16)	0.49 (.21)	0.43	

Semitendinosus -VM Co contraction

Only small differences were seen in the ensemble curves for semitendinosus-VM co-contraction, with slightly lower values of co-contraction for the rocker shoe between 20-45% and 70-85% of stance but higher values from 45%-70% (Figure 6-12). For 15-25% of stance, ANOVA analysis of mean values showed no significant effects of the intervention (Table 6-12).

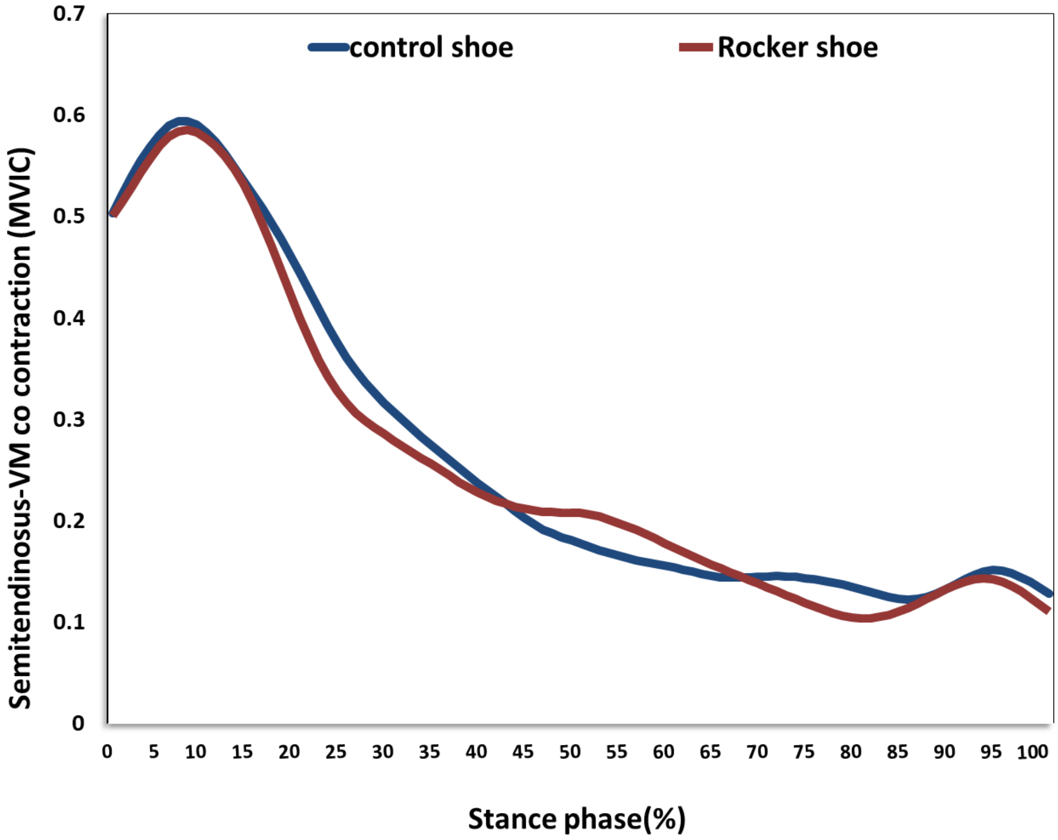

Figure 6-12. Ensemble average of Semitendinosus -VM co contraction in stance phase for knee OA group in different shoes (red-rocker and blue –control shoes).

Table 6-12. Summary result of semitendinosus-VM co-contraction before and after wearing rocker shoes in healthy and knee OA group (15-25%) stance.

semitendinosus-VM co-contraction	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.52 (0.37)	.50 (.39)	0.05	***Effect of shoes=0.42*** ***Effect of group=0.40***
Healthy group	0.45 (0.25)	.41 (.23)	0.16	***Interaction effect=0.73***

Lateral gastrocnemius-VL Co contraction

The ensemble curve for lateral gastrocnemius-VL showed slightly lower co-contraction in the rocker shoe across the middle of stance (Figure 6-13) but minimal differences across the period of interest. Small effects were found from ANOVA analysis across the period 15-25% stance phase: co-contraction was 5% higher for the rocker shoe in the knee OA group (effect size = 0.08), and almost 12% higher for the healthy group (effect size = 0.29). With these small effects, there was no significant difference in co-contraction between the different footwear types (Table 6-13).

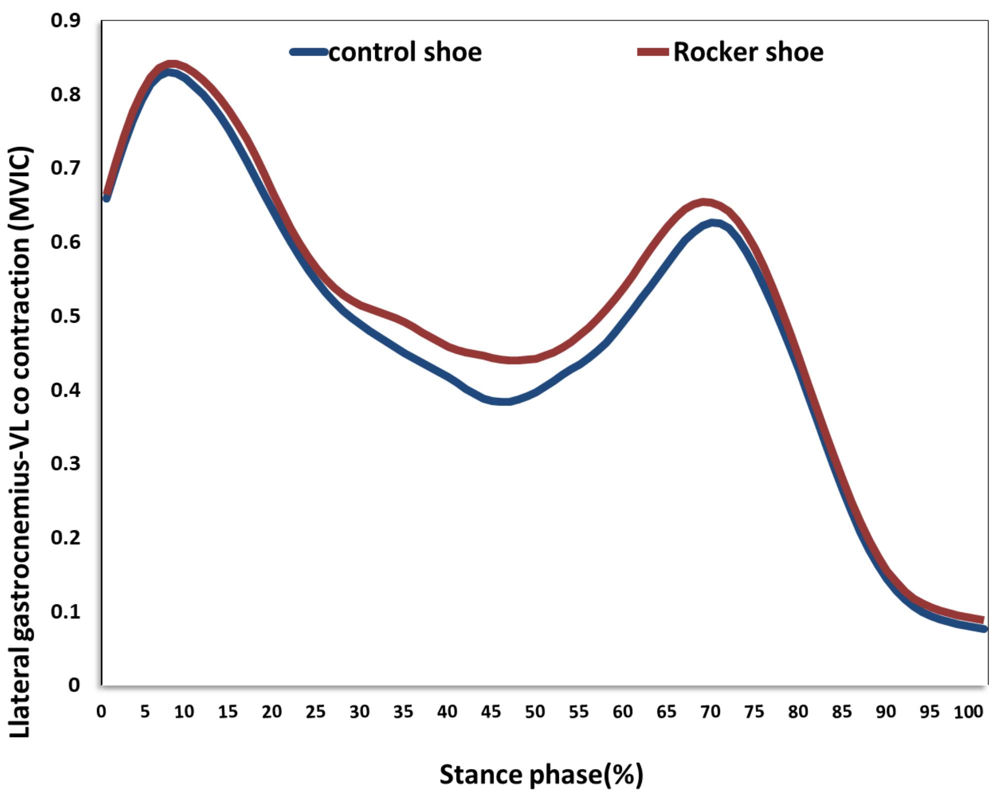

Figure 6-13. Ensemble average of lateral gastrocnemius-VL co contraction in stance phase for knee OA group in different shoes (red-rocker and blue –control shoes).

Table 6-13. Summary result of LG-VL co-contraction before and after wearing rocker shoes in healthy and knee OA group (15-25%) stance.

LG-VL co contraction	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.74 (0.46)	.78 (.45)	0.08	***Effect of shoes=0.17*** ***Effect of group=0.004**** ***Interaction effect=0.85***
Healthy group	0.43 (0.16)	.48 (.18)	0.29	

Medial gastrocnemius -VM Co contraction

Very little difference was observed in the ensemble curves for MG-VM co-contraction for the two shoe types across the stance phase, with only a minimal reduction in co-contraction across the period 20-50% and 60-80% of stance (Figure 6-14). Focusing on 15-25% of stance, ANOVA analysis revealed no effect from shoe, group or interaction (Table 6-14).

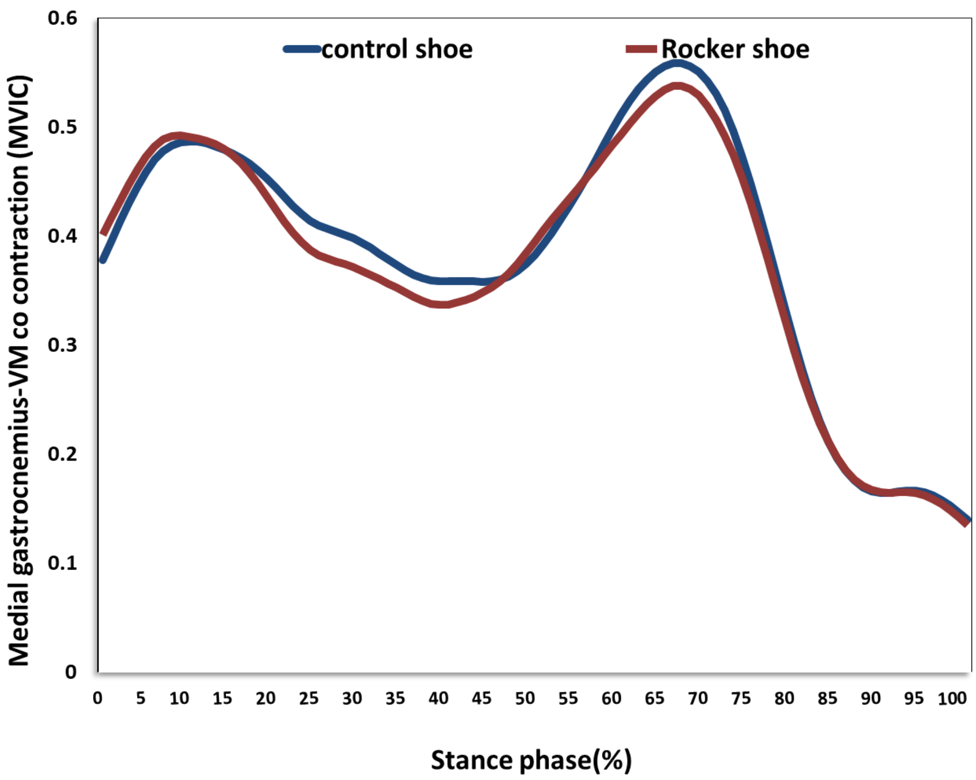

Figure 6-14. Ensemble average of medial gastrocnemius-VM co contraction in stance phase for knee OA group in different shoes (red-rocker and blue –control shoes).

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Table 6-14. Summary result of MG-VM co-contraction before and after wearing rocker shoes in healthy and knee OA group (15-25%) stance

MG-VM co contraction	Control shoe Mean (SD)	Rocker shoe Mean (SD)	Effect size(ES)	*P*
OA group	0.47(0.34)	.48 (.33)	0.02	• ***Effect of shoes=0.80*** • ***Effect of group=0.51*** • ***Interaction effect=0.42***
Healthy group	0.42 (0.22)	.41 (.21)	0.04	

6.4.6 Are there immediate changes in measure of pain when people with knee OA wear a three-curve rocker shoe?

The knee pain in knee OA group was not changed after wearing rocker shoes. The Wilcoxon signed-rank Test -equivalent to the dependent t-test- show no significant difference before and after wearing rocker shoes (Table 6-15)

Table 6-15. Knee pain before and after wearing rocker shoes in the knee OA group during walking.

Knee pain	Mean (SD)
Pre-rocker shoes	1.8 (0.9)
Post- rocker shoes	1.8 (0.9)

6.5 Discussion:

This study aimed to investigate the impact of a 3-curved rocker shoe on trunk inclination, lower limb joint moments, and muscle co-contraction, as well as on reports of pain. The study was based on the idea that an intervention which reduced forward trunk inclination might have effects on the sagittal joint moments and increased co-contraction around the knee seen in knee OA patients. Rocker shoes were selected as an intervention due to their potential to encourage postural realignment, given the instability that they introduce (Sousa & Tavares, 2014). I chose to test the 3-curve rocker shoe design based on its potential for aligning the GRF with the hip, knee and ankle joints across the gait cycle, which, in theory, should reduce the moments about each of these joints (Hutchins et al., 2010), and therefore reduce muscle activity and possibly co-contraction.

The 3-curve rocker shoe was selected from three possible design types, two of which had been previously used in knee OA research: the first was the Masai Barefoot Technology shoe tested by Nigg et al. (2006), who found it to improve static balance and reduce pain; and Tateuchi et al. (2014), who found that it decreased knee flexion moment and increased trunk inclination in the extension direction, suggesting a reduction in forward lean. The second was the Skechers Shape-ups shoe tested by Madden et al. (2015), who found that it decreased peak knee adduction moment, with no change in knee flexion moment. The other possible design was the 3-curve rocker shoe first used by Hutchins et al. (2012), who found lowered muscle activity around both ankle and knee joints using this design. This design was selected for the current study due the possibility that it might lead to reduced co-contraction given the fundamental idea underlying the design which aims to reduce moments at all three joints. Furthermore, in the current study, it was hypothesised that reducing forward trunk lean and hip extensor moment may reduce the excess hamstring activation and hamstring-quadriceps co-contraction reported in knee OA patients during walking (Zeni et al., 2010; Childs et al., 2004; Hortobagyi et al., 2005). Note that this shoe design has been little tested and is novel for knee OA research. This discussion section will first consider the shoe design chosen for the study. It will then discuss the results for each of the four research questions in turn. Then, a summary of the main findings, limitations, conclusions and recommendations will be provided.

6.5.1 Trunk inclination

Trunk inclination is a novel measure for studies of knee OA patients during gait. As expected, the results showed that both knee OA and healthy subjects exhibited mean reductions in forward trunk inclination when wearing the rocker shoe of 30% and 31% respectively, with a pattern which continued across the whole of the stance period. This appeared to confirm the ability of the 3-curve rocker shoe to cause segmental realignment, as reported by Sousa &

Tavares (2014). It is possible that improvements in trunk lean, which were more pronounced in the people with knee OA, were a result of the destabilizing effect of the rocker profile which caused individuals to alter the segmental alignment of their upper body in order to maintain balance on an unstable base. In addition, it is also possible that the observed reductions in forward lean were a direct result of the alteration in the direction of the GRF which is known to occur with a rocker profile (Hutchins et al., 2012).

While there are no other knee OA studies measuring this parameter with footwear interventions, it is interesting to compare with findings from other studies outside the knee OA field. Ochsmann et al. (2016), studying an all-male and healthy cohort, also found significantly reduced trunk inclination in a different rocker design, while Tateuchi et al. (2014) found an impact on trunk lean toward extension direction, i.e. a decrease in forward lean. However, in contrast, Talaty et al. (2016), in a healthy male cohort and different rocker design, report increases in trunk flexion. These contrasting findings, one rocker shoe showing an increase in trunk lean and the other showing a decrease, supports our choice of rocker profile and suggests that further research is required into the possible therapeutic benefit of the three-curved rocker shoe in other clinical populations which may benefit from improvements in upper body posture.

This study was novel in focusing on 15-25% stance (the period of peak loading) in a cohort of individuals with knee osteoarthritis. In addition, the majority of other studies looking at footwear interventions and knee OA have not measured trunk lean, so the results here showing a clear decrease in trunk lean are particularly interesting. In the discussion section of chapter 3, I outlined the possibility that forward trunk lean may be connected to tight hip flexor muscles (Kagaya et al., 2003; Sato & Maitland, 2008). The results presented in this chapter show a reduction in trunk lean when wearing rocker footwear. It is possible that this improvement resulted from a change in the activation of the hip flexor muscles and therefore less muscular restriction during walking. However, I did not measure the hip flexor muscles as part of this investigation and therefore further research is required to confirm this idea.

6.5.2 Lower limb joint moments

Wearing the rocker shoes led to an 8% reduction in sagittal hip moment for the knee OA group and 12% for the healthy group. However, although the reductions in hip moment for the knee OA group extended from 20-70% of stance, the magnitude of this between-shoe difference was not statistically significant. One of the key ideas presented in the literature review of this thesis was that reductions in trunk inclination would lead to reduction in hip moments. However, the data did not fully support this idea. This leaves two possibilities: first, that the changes seen in trunk inclination were not big enough to bring about sufficiently large changes in hip moment; or second, that another mechanism is acting to increase the moments. To explore these ideas further, it is necessary to consider the mechanics of the rocker shoe.

The three-curve rocker shoe design was intended to decrease hip moments, through the introduction of instability, which may encourage realignment of the segments of the body (Sousa & Tavares, 2014). Reductions in trunk inclination were expected to reduce the need for increased hamstring activation to maintain balance, leading to a reduction in hip moment. Hutchins (2012) designed the 3-curve rocker shoe with curves aligned with the sagittal plane centres of the 3 lower limb joints and intended to reduce joint moments, and Buchecker (2013) found a reduction in hip moments using this design. However, in the current study, only a small (non-significant) reduction in the hip extensor moment was observed. This could have been due to the concomitant change in the magnitude and direction of the GRF vector which may have offset the effect the effect of a small decrease in trunk inclination. Alternatively, it is possible that the relatively small change in trunk inclination (1.4° in the knee OA group) was not sufficiently large to elicit a significant change in the hip moment. Interestingly, in their study comparing individuals with a natural forward lean and natural backward lean, Leteneur et al. (2009) observed a mean difference of about 4.6° between the two groups to correspond to a difference of 0.2 Nm/Kg. This compares to a mean difference of 1.4° in lean when wearing rocker and control shoes, with a difference of .03 Nm/Kg in mean hip moment.

When looking at the available literature on footwear interventions in knee OA, a mixed picture is seen regarding the effects of interventions on hip moments. Sobhani et al. (2013), studied a rocker shoe with a proximally positioned profile which differed from the shoe used in this study and found no change in hip moments, which matches the observations in this study. However, in another study, Buchecker's (2013) observed a reduced concentric hip output with an MBT shoe. Overall, it would appear that rocker shoes are only capable of bringing about relatively small changes in hip moments in knee OA populations which, in general, are not significant.

This study observed only small (non-significant) increases in sagittal knee moments in both groups when wearing the 3-curve rocker shoe. However, although the differences over the 15-25% stance period, were small, more pronounced differences were observed in later stance (Figure 6-3). The small differences in knee moment over 15-25% stance suggest that the rocker footwear brought about minimal changes in the direction of the GRF vector during this period. However, later in stance, the internal flexor moment was reduced dramatically and became an internal extensor moment. These data demonstrate a large shift in the direction of the GRF vector relative to the knee joint centre during this phase of the gait cycle and therefore a substantial change in the moment. Nevertheless, as knee moments are typically small during this phase of gait cycle, it is unlikely that change would results in a substantial change in knee joint loading.

Previous studies into the effects of rocker footwear on knee moments show mixed results. For example, two studies observed rocker footwear to bring about a reduced external knee flexor moment during loading response (Buchecker et al., 2013, Tateuchi et al., 2014a). However, in a study of MBT rocker shoes, Sobhani et al. (2013) found no change in the knee flexion moment in early stance. These latter results are consistent with the findings of this study. Sacco et al. (2012) state that their results with an MBT shoe, in which vertical loads were increased in comparison with standard shoes and walking barefoot, suggest increased musculoskeletal loading when the shoe was first worn. Overall, the research does not point to a clear and consistent effect of rocker footwear on knee moments and this is most likely the result of variations in the outsole profile design of the different shoes shoe having a different effect on the change in the direction of the GRF vector and possibly the position of the upper body.

The data presented in the section above showed a clear effect of the rocker footwear on the ankle moments over the period of interest (15-25% stance) for both groups. This difference appeared to be maintained throughout midstance: however, I did not perform any statistical analysis outside the period of interest and so it is not clear whether these differences are significant. Nevertheless, Figure 6-4, shows a large effect of the footwear, illustrated by a large shift in the curve relative to the standard deviation band. Hutchins et al. (2012) suggest that the three-curve rocker shoe will alter the direction and orientation of the GRF during walking in such a way as to redirect the GRF so that it is closer to the ankle joint centre during early stance. My results support this idea and suggest that this redirection of the GRF occurred from 10-60% stance. This idea is consistent with the results of other studies into rocker footwear (Taniguchi et al., 2012, Boyer and Andriacchi, 2009, Hutchins et al., 2009) which have reported ankle plantarflexor moment during walking. In general, these studies demonstrate reduced plantarflexor moments over early to mid-stance and this effect appears to occur irrespective of the precise rocker profile.

6.5.3 EMG muscle activity

The results for EMG assessment of the hamstrings showed minimal difference in the biceps femoris over the period of interest but a consistent (however non-significant reduction) in semitendinosus activity across the gait cycle (Figure 6-6). Interestingly, although there was no difference in biceps femoris across early stance, wearing the rocker shoe did seem to be associated with a reduction in activity during midstance (Figure 6-5). The lack of an effect of the rocker shoe on hamstring activity fits with the data, presented above, showing only small reductions in hip extension moment. Given these small changes in hip moment, it would be reasonable to anticipate only small reductions in hamstring activity. However, larger change in hamstring activity later in midstance is difficult to explain and could possibly be the result of an increase in gluteus maximus activation, a synergist for hip extension.

It is interesting to compare these findings on hamstring activity with those from previous rocker shoe research. A study on MBT rocker footwear worn by healthy subjects conducted by Sacco et al. (2012) reported no differences in EMG results between the rocker shoe and a control shoe. Further, Santo et al. (2012) found no change in biceps femoris muscle activity when walking in rocker footwear. Likewise, Forghany et al. (2014) found no significant effects of MBT or rollover shoes on biceps femoris activity. The results of the current study are therefore comparable to previous literature on this measure and confirm the idea that rocker footwear is unlikely to reduce hamstring activity, irrespective of the rocker profile.

The EMG results for the quadriceps also showed no clear increase in activity across the period of interest. Again, given the minimal changes in the knee moment across this period, this lack of a change in the EMG data would be expected. However, there did appear to be a slight increase in vastus medialis activity during midstance (Figure 6-7). This may indicate that the instability resulting from the rocker shoe could lead to increased activity in the quadriceps.

However, further analysis across the midstance period would be needed to test this idea but this was deemed outside the scope of the current study.

Considering previous studies, Sacco et al. (2012) did not find any significant difference in EMG activity for the vastus lateralis between MBT rocker and standard footwear for a healthy cohort. Moreover, Tan et al. (2016) found no significant effects of MBT rocker shoes on quadriceps activity in their systematic review. This present study is therefore in line with previous work on other rocker shoes in finding no significant link with quadriceps activity.

The EMG results for the gastrocnemius muscles show a decrease for the medial gastrocnemius particularly from 30-75% of stance, but also a slight decrease between 10 and 30%. However, the strength of these changes is not significant, and little change is seen for the lateral gastrocnemius muscle in relation to the rocker shoe. Overall, these results do not match the reduction in sagittal ankle moment which is seen in the period of interest.

Santo et al. (2012) did not find alterations in gastrocnemius activity in rocker shoes, similar to this study. However, Forghany et al. (2014) found increased medial gastrocnemius activity, which, while not statistically significant, contrasts with the decrease observed in this study. Tan et al. (2016) found no significant alterations in amplitude or time of gastrocnemius muscle activity when reviewing trials of MBT rocker shoes. Overall therefore, the results here fit within the wider literature in finding little change in the activity of these muscles from rocker footwear.

6.5.4 Co-contraction

The data comparing co-contraction for the four muscle pairs (Figure 6-11 – 6-14) showed minimal differences between the control and the rocker shoe across the 15-25% period of interest. During this time, no significant changes were identified in any of the four measures of co-contraction. These results are logical when viewed alongside the muscle activity results for this period, which also showed minimal changes in EMG patterns with the rocker footwear. This result is important as it shows that rocker footwear is unlikely to decrease co-contraction in people with knee OA. As explained in the literature review and demonstrated in Chapter 4, people with OA have a tendency to walk with increased co-contraction of the knee muscles (Childs et al., 2003), which may cause the knee joint to deteriorate (Lewek et al., 2005). These OA-type muscle activation and co-contraction patterns may result in elevated joint contact force at the medial compartment (Sritharan et al., 2016a). The data from this study does not support the idea that the three-curve rocker shoe could reduce co-contraction and therefore does not support is use clinically in knee OA populations.

It is interesting to compare the data in this study with previous studies which have investigated co-contraction in rocker footwear. Horsak et al. (2015) found a statistically significant increase in co-contraction between BF and vastus medialis across the gait cycle. Although I focused on a specific phase of the gait cycle, my data do not seem to be consistent with this finding which may possibly be because of the different rocker profile used by Horsak. However, the idea of rocker shoe increasing co-contraction is also supported by the data of Buchecker et al. (2010), who noted that vastus lateralis activity, as well as gastrocnemius activity, was greater in a rocker shoe in the late phase of stance, and that this lead to an increased co-activation between these muscles at mid- and end-stance. These previous studies suggest a destabilising effect from the rocker shoe which leads to increased co-contraction. However, I did not observe this effect and so it would appear that alterations in co-contraction are dependent on rocker outsole profile.

6.5.5 Effect of 3-curve rocker shoe on pain

The investigation of pain experienced by the knee OA group when walking in control shoes or in the three-curve rocker shoe found no immediate changes. This is unsurprising given the findings discussed above with regard to the lack of a change of co-contraction of the lower limb muscles. Nigg et al. (2006) studies the Masai Barefoot Technology (MBT) shoe and looked at clinical outcomes relating to pain and function. They and found significant improvement in static balance for the MBT shoe which was accompanied by reductions in pain. The authors concluded that rocker shoes may be effective in reducing pain in people with knee OA and suggested a mechanism focused on the idea that the rocker shoe reduced the force exerted between bones, redistributed body weight and changed the activity of the muscles surrounding the joint (Nigg et al., 2006). In the current study, there was no immediate change in pain before and after wearing rocker shoes. One explanation for this is that the three-curve rocker shoe did not reduce co-contraction. However, alternatively, it is possible that the short time frame of the experimental trials was insufficient to produce any marked reduction in reported pain.

6.5.6 Clinical implications

We hypothesised that the three-curve rocker shoe would lead to a segmental realignment (more upright position of the trunk) during walking. It was thought that if this was achieved, it might lead to a corresponding reduction in sagittal hip moments and muscle co-contraction and that this could reduce the articular load, thereby improving pain. However, while the rocker shoe had the effect of reducing forward trunk inclination, minimal changes were observed in co-contraction and there was no change in reported pain. Therefore, the three-curve rocker shoe may not be a suitable clinical intervention for improving gait in people with knee OA.

6.5.7 Limitations of the study

As explained in the previous chapter of this thesis, the measurement of trunk lean is problematic, due to the complex and flexible structure of the spine. Trunk lean was quantified by using a single thoracic segment which was tracked using four markers attached at fixed points and defined using markers on the acromiums and the greater trochanters. However, whereas in the previous study (chapter 5), I was comparing trunk inclination across different individuals, in this present study I was looking for within-subject changes in trunk inclination. Using a within-subject design, reduces the potential for error which results from inter-subject differences in trunk motions. However, the limitation of using a single rigid body model of the trunk still remains and may have introduced a degree of uncertainty into the trunk inclination measurements.

A further limitation of the study comes from the use of surrogate measures (EMG and joint moments) to gain insight into joint loading. It is possible that the rocker footwear did result in reductions in joint loading, but I was unable to capture this with my measurement. To obtain more precise estimates of joint loading, it is necessary to develop a full musculoskeletal and finite element model. This was deemed to beyond the scope of this thesis. Nevertheless, I used a measure of co-contraction which has been shown to have a reasonable correlation with joint load (Brandon et al., 2014, Sritharan et al., 2016a). This measure was focused on a time window 15-25% stance, a choice which was justified from interpretation of modelling studies showing the effect of increased co-contraction on peak knee forces (See section 2.4.5 for more details). With this outcome, the analysis provided insight into whether rocker footwear may have influence joint loading. However, it is possible that this narrow statistical focus did not capture other changes in muscle patterns which may have resulted from the rocker footwear.

Nevertheless, this decision was made to minimise the chance of type 1 error, which would have increased if additional outcomes (focusing on additional time windows) had been analysed.

A single, specific rocker outsole profile was chosen for the study, and this presents a limitation, as it is not clear whether or not findings would have been similar using other designs of rocker footwear. This limitation was made necessary by the practical limitations of the study, including time and resources. While other studies have focused on different rocker shoe designs, including MBT and Skechers Shape-ups, the 3-curve design used for this study was chosen based on data and the suggestion by Hutchins (2012) that it could reduce joint moments. Nevertheless, most of the patterns that we observed were generally consistent with previous studies and suggest that rocker footwear is unlikely to result in large changes to muscle co-contraction.

6.6 Conclusions

The three-curve rocker shoe led to a reduction in forward trunk inclination, as hypothesised. Howeer, this was not accompanied by reduced joint moments, reductions in hamstring activity or reduced co-contraction. It is likely that the reductions in trunk inclination were insufficient to bring about the changes in hip moments and muscle patterns that were hypothesised. Therefore, as reductions in moments and in co-contraction were not observed, the results suggest that rocker footwear may not be a viable intervention for people with knee OA if the aim is to reduce sagittal plane loads.

Chapter 7 - Final conclusions and future recommendations

7.1 Introduction

This thesis has extended the knowledge base related to gait in knee OA patients, and in particular has contributed to the understanding of forward trunk inclination for this group, through three linked studies. It has explored a new model to explain altered moment and muscle activity in people with knee OA, based on sagittal plane trunk inclination altering the direction of the ground reaction force vector and therefore the moments and muscle activation patterns at the hip knee and ankle. The data show a clear pattern of increased forward lean in people with knee OA while walking and confirms previous research showing that people with knee OA also stand with increased forward lean (Turcot et al., 2015). The data presented in this thesis also supports previous work which shows differences in lower limb joint moments, muscle activity and co-contraction between people with knee OA and healthy subjects. However, in contrast to much earlier research which has often focused on peak values (Brandon et al., 2014, Sritharan et al., 2016a), I focused on the period (15-25% of stance) which corresponds to the point of peak loading. In chapter 5, I investigated the links between a range of biomechanical parameters and trunk inclination and found weak-moderate correlations between trunk inclination and hip moments/muscle activations. Although this link was less clear that I originally hypothesised, these data do provide new insights into the gait mechanics of people with knee OA and motivate further study in this area. The thesis has also extended understandings of rocker shoe interventions for knee OA, finding reductions in forward lean from a 3-curved rocker shoe, but also showing minimal changes in other biomechanical variables over the period of interest (15-25% stance).

The final chapter of the thesis will review the rationale for the project, its objectives, and the outcomes of the three studies conducted in comparison to the overall aim of the thesis and the objectives set for each study. Further, it will synthesise results to give an overview of what has been learnt in terms of the overall contribution to the body of knowledge. The limitations of the project will then be presented, before the extendibility of the study is discussed and recommendations for future directions made.

7.2 Overview of results

First, the literature related to gait and knee OA was reviewed and synthesised, and this resulted in several main findings on important characteristics of gait in knee OA:

- Moments: peak knee extensor moments are reduced but hip extensor moments increased during midstance in people with knee OA.
- Muscle activity: hamstring activity, quadriceps activity and gastrocnemius activity are increased in people with knee OA.
- Joint loading: the elevated levels of muscle activity lead to increased compressive loading at the knee joint at between 15-25% stance phase.

While these main features are generally accepted however, explanations for these alterations vary, with one perspective viewing them as an appropriate strategy to stabilise the knee joint in knee OA, while another perspective views the responses as maladaptive, increasing compression at the knee joint and therefore speeding the progression of the disease.

The literature review also formed a basis for the formulation of a hypothesis to explain some of the modifications to gait seen in knee OA. Briefly, it was proposed that increased hip extension moments, decreased knee extension moments and increased muscular co-contraction during 15-25% of stance phase (Liu et al., 2014) may result from increased forward trunk inclination, which has been found to have similar effects in studies of gait more generally, because of the need to support the trunk against gravity (Leteneur et al., 2009). This was considered to result from anterior shifting of the ground reaction force vector relative to the hip joint and knee joint in response to the altered centre of mass from forward trunk lean, thereby increasing hip extensor moment and decreasing knee moment, as reported by Huang et al. (2008) and Debbi et al. (2014) respectively. The hamstrings-quadriceps co-contraction increase seen in knee OA (Zeni et al., 2010; Childs et al., 2004; Hortobagyi et al., 2005) was hypothesised to be linked with increased hip extensor moment from trunk inclination, as hamstrings are responsible for extending the hip. Further, the potential was identified for increased gastrocnemius-quadriceps co-contraction resulting from increased trunk inclination if shifting A-P CoP and therefore ankle moment is changed by this. In investigating these possibilities, the thesis has presented an original approach to gait knee OA. There is previous work investigating the links between trunk position in the frontal plane and knee adduction moments, but little exploration of the characteristics and consequences of sagittal plane trunk alignment in people with knee OA.

7.2.1 Study One: Trunk inclination in people with knee OA

The first of the three studies explored differences in sagittal plane trunk inclination between healthy people and individuals affected by knee OA. It first explored trunk inclination during standing and walking through the following research questions:

RQ 1A: Do individuals with knee OA walk with an increased inclination of the trunk?

RQ 1B: Do individuals with knee OA stand with an increased inclination of the trunk?

RQ 1C: Does trunk inclination in standing correlate with trunk inclination in walking both in a group of individuals with knee OA and also in a healthy cohort?

Following this, CoP was investigated to find potential links to forward trunk inclination:

RQ 1D: Is there a difference in CoP between healthy and knee OA subjects?

RQ 1E: Is there a link between forward trunk inclination and anterior shift of CoP?

Finally, moments, muscle activation and co-contraction were compared for healthy people and those with knee OA:

RQ 1F: What are the differences in hip/knee/ankle moments between healthy and knee OA subjects?

RQ 1G: What are the differences in hamstring/quadriceps/gastrocnemius muscle activity between healthy and knee OA subjects?

RQ 1H: What are the differences in the co-contraction between healthy and knee OA subjects?

The main findings from this study support the differences in gait in people with knee OA from previous studies which were summarised in Section 7.2, concerning moments and muscle activity. In particular, the knee OA group showed muscle activations in biceps femoris activity, semitendinosus activity and vastus lateralis activity which were typically 50% larger than in the healthy people._Importantly, the findings also reveal significant differences between the OA and control groups in terms of standing and walking of sagittal trunk inclination, with mean increased forward lean for the knee OA group of 1.7° while standing and 3° during walking. However, only a weak correlation (r=0.42) between standing and walking was identified, suggesting a different mechanism for inclination in standing and in walking. Two mechanisms were considered in this thesis for forward trunk lean in walking. First was the theory of realignment through anterior displacement of CoP to maintain balance, which would show increases in the ankle plantar flexor moment due to the anterior shift on the CoP relative to the ankle joint. However, this appeared to be unlikely given the lack of an association between trunk inclination and change in CoP and no corresponding change in ankle moment. The second theory was that tightness in hip flexor muscles had a causal role in forward lean, and the findings were in line with this concept. While I did not collect data to fully support this idea, it appeared possible and therefore further research is required in this area.

7.2.2 Study Two: The relationship between trunk inclination and joint moments/muscular co-contraction

The second study investigated links between trunk inclination and lower limb moments/muscle activation and co contraction patterns in the sagittal plane while walking, to understand more fully the mechanism for forward lean. The study therefore examined a novel concept in knee OA and a comparatively under-researched area of gait analysis more generally. This was approached through three research questions:

> RQ 2A: What is the relationship between trunk inclination and hip/knee/ankle moments in people with knee OA and also in healthy control subjects?
>
> RQ 2B: What is the relationship between trunk inclination and hamstring/quadriceps/gastrocnemius activity in people with knee OA and also in healthy control subjects?
>
> RQ 2C: What is the relationship between trunk inclination and co-contraction in people with knee OA and also in healthy control subjects?

The main findings of study two include a weak to moderate association between forward trunk inclination and sagittal hip moment and weak correlation between trunk lean and hamstring muscle activation. Although these findings are consistent with some previous research on healthy participants (Kluger et al., 2014; Leteneur et al., 2009; Sato & Maitland, 2008), the correlations were weaker than anticipated. Nevertheless, the data do provide support for the idea that increased trunk inclination could lead to increase hip moments and therefore hamstring activation patterns. However, it is possible that uncertainty in trunk measurements which results from anatomical variation, along with other associated uncertainties in hip kinetic/EMG measurement could have contributed to the lower-than-anticipated correlations.

7.2.3 Study Three: The biomechanical effects of rocker footwear in people with knee OA

The final study in the thesis considered a footwear intervention for knee OA in relation to the aspects of gait studied in the first two studies. Based on literature, a 3-curved rocker shoe was identified as a potential intervention to influence sagittal joint angle, moments and upper body positioning, and in particular to alter sagittal plane alignment of the trunk and produce corresponding alterations in joint moments and patterns of muscle activation. While previous literature has trialled various designs of footwear as a knee OA intervention, this is the first to focus on this specific aim. This was done through the following research questions:

> RQ 3A: How does inclination change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?
>
> RQ 3B: How do lower limb moments change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?
>
> RQ 3C: How does muscular co-contraction change when people with knee OA or healthy control subjects wear a three-curve rocker shoe?
>
> RQ 3D: Are there immediate changes in measure of pain when people with knee OA wear a three-curve rocker shoe?

It was anticipated that the intervention would reduce forward trunk inclination, and this was clearly observed, with an average 1.4° reduction for the knee OA group across the gait cycle. However, this was not accompanied by the expected changes in joint moment or hamstrings activity. Further, there was no immediate change noted in the pain reported by participants. Based on these findings, it is possible that trunk inclination was insufficiently altered by the intervention to reach the desired outcomes for the other parameters. Alternatively, this specific design of rocker shoe may have had other effects which were not anticipated, such as destabilisation effects, which counteracted any effects from the change in trunk inclination and produced the unexpected results. The findings produced do not point to the viability of the 3-curved rocker shoe as an intervention for knee OA.

7.3 Clinical Recommendations

Overall, the thesis demonstrates an increased forward trunk lean in people with knee OA. Therefore, strategies to decrease this may be warranted in clinical practice. These may include therapeutic approaches such as a programme of tailored stretching exercises, as well as postural re-education approaches and/or muscle strengthening programmes. Clinical programmes could also include proprioceptive training in body position awareness and/or biofeedback training to re-educate patients on the correct position of the trunk during walking.

In the second study, a weak, but significant, correlation was observed between trunk inclination and hamstring muscle activity. Given that increased muscle activity is associated with increased compressive force at the joint (Andriacchi, 1994, Zeni et al., 2010), then approaches which could reduce trunk inclination, such as those described above, could have an impact on muscle patterns, thereby reducing contact loading at the joint. However, further research is required to fully validate this idea.

The findings of the final study do not support the use of 3-curve rocker shoe as a footwear intervention. This is because, although the shoe reduced forward trunk inclination, it did not have corresponding effects, reducing either hamstrings activity or joint moments. This suggests that this type of footwear may not be beneficial for reducing joint loading and therefore not appropriate for people with knee OA. Further research is therefore needed to understand the potential effectiveness of other types of rocker footwear could be a better alternative treatment if the target is to reduce muscle activity.

7.4 Recommendations for future research

The outcomes of this study advance knowledge in relation to walking in knee OA, and particularly in terms of trunk inclination: however, they also raise questions which could usefully be explored further. Future research recommendations are therefore given as follows:

I proposed that tightness in hip flexor muscles may acts as a mechanism for forward trunk inclination. However, further evidence is needed to fully validate this idea, particularly in the area of knee OA. Such studies could be both cross sectional in nature, investigating correlations between natural trunk inclination and hip flexor muscle length or intervention studies, investigatintg the effects of stretching hip flexor muscles on upper body position.

This results of this thesis also motivates future study into the effects of biofeedback training on trunk inclination. With recent advances in motion capture technology, it is now possible to give participants real-time feedback on the position of their trunk. Using this appraoch, it would be possible to precisely quantify the effect of small, imposed changes in trunk inclination on joint moments, muscle activations and co-contraction. Such an approach would reduce the uncertainty associated with comparing trunk inclination across different people and may provide improved insight into th eeffect of trunk inclination on gait parameters related to joint loading.

Finally, more research is needed into the full effects of rocker shoe designs in people with knee OA. I observed some interesting effects in which trunk inclination was reduced but co-contraction and hamstring activity did not decrease. Future studies may determine how different rocker shoe profiles may impact on specific gait parameter, related to knee loading, and establish whether optimised footwear designs could be used as a treatment option for people with knee OA.

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List of appendices

Appendix 1: University of Salford ethical approval

University of Salford
MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

College of Health & Social Care
AD 101 Allerton Building
University of Salford
M6 6PU

T +44(0)161 295 2280
HSresearch@salford.ac.uk

www.salford.ac.uk/

26 June 2015

Dear Ali Saad,

RE: ETHICS APPLICATION HSCR 15-35 – The biomechanical effect of rocker shoes in people with knee osteoarthritis

Based on the information you provided, I am pleased to inform you that application HSCR15-35 has been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible by contacting HSresearch@salford.ac.uk

Yours sincerely,

Appendix 2: NHS ethic approval

NHS
Health Research Authority

Dr Stephen John Preece
Research Fellow
University of Salford
Blatchford Building
University of Salford
Manchester
M6 6PU

Email: hra.approval@nhs.net

22 February 2017

Dear Dr Preece

Letter of HRA Approval for a study processed through pre-HRA Approval systems

Study title:	**The biomechanical effect of rocker shoes in people with knee osteoarthritis.**
IRAS project ID:	**186778**
Sponsor	**University of Salford**

Thank you for your request for HRA Approval to be issued for the above referenced study.

I am pleased to confirm that the study has been given HRA Approval. This has been issued on the basis that the study is compliant with the UK wide standards for research in the NHS.

The extension of HRA Approval to this study on this basis allows the sponsor and participating NHS organisations in England to set-up the study in accordance with HRA Approval processes, with decisions on study set-up being taken on the basis of capacity and capability alone.

If you have submitted an amendment to the HRA between 23 March 2016 and the date of this letter, this letter incorporates the HRA Approval for that amendment, which may be implemented in accordance with the amendment categorisation email (e.g. not prior to REC Favourable Opinion, MHRA Clinical Trial Authorisation etc., as applicable). If the submitted amendment included the addition of a new NHS organisation in England, the addition of the new NHS organisation is also approved and should be set up in accordance with HRA Approval processes (e.g. the organisation should be invited to assess and arrange its capacity and capability to deliver the study and confirm once it is ready to do so).

Please note that this specifically includes the amendment submitted on 5 December 2016 which seeks approval to involve local GP practices.

Participation of NHS Organisations in England

Please note that full information to enable set up of participating NHS organisations in England is not provided in this letter, on the basis that activities to set up these NHS organisations is likely to be underway already.

The sponsor should provide a copy of this letter, together with the local document package and a list of the documents provided, to participating NHS organisations in England that are being set up in accordance with HRA Approval Processes. It is for the sponsor to ensure that any documents provided to participating organisations are the current, approved documents.

For non-commercial studies the local document package should include an appropriate Statement of Activities and HRA Schedule of Events. The sponsor should also provide the template agreement to be used in the study, where the sponsor is using an agreement in addition to the Statement of Activities. Participating NHS organisations in England should be aware that the Statement of Activities and HRA Schedule of Events for this study have not been assessed and validated by the HRA. Any changes that are appropriate to the content of the Statement of Activities and HRA Schedule of Events should be agreed in a pragmatic fashion as part of the process of assessing, arranging and confirming capacity and capability to deliver the study. If subsequent NHS organisations in England are added, an amendment should be submitted to the HRA.

For commercial studies the local document package should include a validated industry costing template and the template agreement to be used with participating NHS organisations in England.

It is critical that you involve both the research management function (e.g. R&D office and, if the study is on the NIHR portfolio, the LCRN) supporting each organisation and the local research team (where there is one) in setting up your study. Contact details and further information about working with the research management function for each organisation can be accessed from www.hra.nhs.uk/hra-approval.

After HRA Approval

In addition to the document, *"After Ethical Review – guidance for sponsors and investigators"*, issued with your REC Favourable Opinion, please note the following:

- HRA Approval applies for the duration of your REC favourable opinion, unless otherwise notified in writing by the HRA.
- Substantial amendments should be submitted directly to the Research Ethics Committee, as detailed in the *After Ethical Review* document. Non-substantial amendments should be submitted for review by the HRA using the form provided on the HRA website, and emailed to hra.amendments@nhs.net.

- The HRA will categorise amendments (substantial and non-substantial) and issue confirmation of continued HRA Approval. Further details can be found on the HRA website.

Scope

HRA Approval provides an approval for research involving patients or staff in NHS organisations in England.

If your study involves NHS organisations in other countries in the UK, please contact the relevant national coordinating functions for support and advice. Further information can be found at http://www.hra.nhs.uk/resources/applying-for-reviews/nhs-hsc-rd-review/.

If there are participating non-NHS organisations, local agreement should be obtained in accordance with the procedures of the local participating non-NHS organisation.

User Feedback

The Health Research Authority is continually striving to provide a high quality service to all applicants and sponsors. You are invited to give your view of the service you have received and the application procedure. If you wish to make your views known please email the HRA at hra.approval@nhs.net. Additionally, one of our staff would be happy to call and discuss your experience of HRA Approval.

HRA Training

We are pleased to welcome researchers and research management staff at our training days – see details at http://www.hra.nhs.uk/hra-training/.

If you have any queries about the issue of this letter please, in the first instance, see the further information provided in the question and answer document on the HRA website.

Your IRAS project ID is 186778. Please quote this on all correspondence.

Yours sincerely

Kevin Ahmed
Assessor

Telephone: 0207 104 8171
Email: hra.approval@nhs.net

Copy to: *Mrs Kay Hack, Sponsor Contact, University of Salford*
Mr Ali Algarni, Student, University of Salford

Appendix 3: Consent form

University of Salford MANCHESTER

Consent form *Version v6 (18.11.15)* ***Subject No:*...............**

Research Participant Consent Form

Title of Project: Rocker shoes in Knee Osteoarthritis

Ethics Ref No: HSCR15/35

Name of Researcher: Ali Algarni

(Please initial box)

- I confirm that I have read and understood the information sheet titled "The biomechanical effect of rocker shoes in people with knee osteoarthritis, v5 (16th November 2015) and what my contribution will be. ☐
- I have been given the opportunity to ask questions (face to face, via telephone and e-mail) ☐
- I agree to digital images being taken during the research exercises ☐
- I understand that my participation is voluntary and that I can withdraw from the research at any time **without giving any reason** ☐
- I understand how the researcher will use my responses, who will see them and how the data will be stored. ☐
- **I agree to take part in the above study** ☐

Name of participant ....

Signature ....

Date ....

Name of researcher taking consent **Ali Algarni**

Researcher's e-mail address : **A.S.S.Algarni@edu.salford.ac.uk**

Adapted from College Ethical Approval Panel for Research Consent Form

Appendix 4: Participant Information Sheet - Healthy subject

Participant Information Sheet

The biomechanical effect of rocker shoes in people with knee osteoarthritis.

INFORMATION ABOUT THIS DOCUMENT

You are being invited to take part in a research study, as a healthy volunteer, to help us understand the possible effects of rocker footwear on the biomechanics of walking. Before you decide, it is important for you to understand why the research is being done and what it will involve. This document gives you important information about the purpose, risks, and benefits of participating in the study. Please take time to read the following information carefully. If you have any questions then feel free to contact the researcher whose details are given at the end of the document. Take time to decide whether or not you wish to take part.

BACKGROUND TO THE STUDY

Individuals with knee osteoarthritis suffer from pain during normal activities such as walking, standing or climbing stairs. We aim to gain more information about the effect of using conservative interventions, such as footwear on the biomechanics of gait and knee joint loading. A number of footwear have been designed to potentially lower the loads in the knee joint and these treatments could be extremely popular, effective and inexpensive interventions for this disease if we can understand which one has the best results. As well as understanding the effect of these footwear designs on people with knee arthritis, it is important to understand how they could change the way healthy volunteers walk.

WHAT WILL HAPPEN TO ME IF I PARTICIPATE IN THIS STUDY?

If you decide that you would like to take part in the study, please contact the researcher on the numbers at the end of this information leaflet. The principal investigator will contact you to ask you a few questions to confirm that you are suitable for the study and answer any further questions you may have.

If you are happy to take part in the study, you will be required to visit the gait laboratory at the University of Salford on a single occasion. At the start of this visit, the study will be explained in full and, if you are happy to proceed, you will complete a consent form. Next, we will measure your height and weight and then ask you to change into shorts and a comfortable T-shirt.

To begin with the researcher will place small electrodes over specific muscles on the front and back of your thigh, on your calf and on the front of your shin. Before each of these electrodes is positioned, the researcher will remove any excess hair with a disposable razor and then remove dead skin with an exfoliating cream. Once the electrodes are in place, the researcher will then position reflective markers on your legs, feet, arms and upper body using hypo-allergenic tape as shown in the picture to the right.

With the electrodes and markers in place the researcher is able to capture biomechanical data during movement. To begin with you will be asked to stand still whilst a standing trial is recorded. We will then ask you to walk in a pair of normal shoes 5-10 times, over a distance of 6 metres, at your normal walking speed. We will then ask you to change into a pairof rocker shoes and to walk another 5-10 times, again at your normal walking speed. Finally, you will change into a pair of flexible shoes and again repeat the 5-10 walks. Note that the rocker shoes have a curved bottom and are designed to ‘rock’ the foot forward thereby making walking a little easier (see picture to the right).If you get tired during any point then we will give you sufficient time to rest. We anticipate that the total duration of this visit would be no longer than 1.5-2 hours.

If you are able to make your own way to the university then you will receive a £25 payment to cover travel expenses. Alternative, we can arrange a taxi to pick you up and to take you back home again. If you do need a taxi, then we will pay for it but we will be unable to provide you with personal expenses payment.

RISKS & POTENTIAL BENEFITS OF THE STUDY

What risks are involved in participating in the study?
This is a very simple, straight forward study with negligible risks. The laboratory measurements of walking are often carried out in routine clinical practice and will be performed by a fully trained researcher with state-of-the are equipment

What benefits are involved in participating in the study?
There are no immediate benefits to you of participating in the study. However, the results will help us understand the effect of footwear interventions on the biomechanics of gait in people with osteoarthritis. This could ultimately help us to develop effective treatments for people who suffer with knee arthritis.

WHAT IF SOMETHING GOES WRONG
The university has insurance to cover against any harm to you which may occur whilst you are taking part in these tests. However, if you decide to take legal action, you may have to pay for this. If you wish to complain, or have any concerns about any aspect of the way you have been approached or treated during the course of this study, you can approach the University of Salford as describe below:

Contact the Research & Innovation Manager:
Mr. Anish Kurien MBA, PRINCE2, MSP
Email: a.kurien@salford.ac.uk, Tel:+44 (0) 161 295 5276

ENDING THE STUDY

What if I want to leave the study early?
You can withdraw from this study at any time without loss of any non-study related benefits to which you would have been entitled before participating in the study. There is no danger to you if you leave the study early. If you want to withdraw you may do so by notifying the study representative listed in the "Contact Information" section below. Moreover all data collected from you will be destroyed, including any personal information.

FINANCIAL INFORMATION

Who is organizing and funding the research?
This study is organized and funded by the University of Salford.

Will I be paid for participating?
We are able to provide a £25 payment to cover your travel expenses to and from the university or, alternatively, to provide you with a taxi.

CONFIDENTIALITY OF SUBJECT RECORDS

Will my taking part in this study be kept confidential?
All information which is collected about you during the course of the research will be kept strictly confidential. Any information about you which leaves the University of Salford will have your name and address and any other identifying features removed so that you cannot be recognized from it.

USE OF THE DATA

The data collected as part of this study will be used to understand the effect of different footwear designs and this will be published both in a Ph.D. thesis and also in scientific journal papers.

CONTACT INFORMATION

If you require more information about the study, want to participate, or if you are already participating and want to withdraw, please contact

	Ali Algarni
Email:	A.S.S.Algarni@edu.salford.ac.uk
Phone :	0161 295 2017
Address:	School of Health Sciences Brian Blatchford Building, University of Salford Salford Manchester M6 6PU

Thank you very much for taking time to read this document!

We appreciate your interest in this study and hope to welcome you at the School of Health Sciences, University of Salford.

Appendix 5: Participant information sheet -OA participants

Participant Information Sheet

The biomechanical effect of rocker shoes in people with knee osteoarthritis.

INFORMATION ABOUT THIS DOCUMENT

You are being invited to take part in a research study to help us understand if a new treatment for knee osteoarthritis could be effective and also what effect it may have on healthy volunteers. Before you decide, it is important for you to understand why the research is being done and what it will involve. This document gives you important information about the purpose, risks, and benefits of participating in the study. Please take time to read the following information carefully. If you have any questions then feel free to contact the researcher whose details are given at the end of the document. Take time to decide whether or not you wish to take part.

BACKGROUND TO THE STUDY

Individuals with knee osteoarthritis suffer from pain during normal activities such as walking, standing or climbing stairs. We aim to gain more information about the effect of using conservative interventions, such as footwear on the biomechanics of gait and knee joint loading. A number of footwear have been designed to potentially lower the loads in the knee joint and these treatments could be extremely popular, effective and inexpensive interventions for this disease if we can understand which one has the best results. As well as understanding the effect of these footwear designs on people with knee arthritis, it is important to understand how they could change the way healthy volunteers walk.

WHAT WILL HAPPEN TO ME IF I PARTICIPATE IN THIS STUDY?

If you decide that you would like to take part in the study, please contact the researcher on the numbers at the end of this information leaflet. The principal investigator will contact you to ask you a few questions to confirm that you are suitable for the study and answer any further questions you may have.

If you are happy to take part in the study, you will be required to visit the gait laboratory at the University of Salford on a single occasion. At the start of this visit, the study will be explained in full and, if you are happy to proceed, you will complete a consent form. Then we will ask you to complete a data access form to give us permission to contact your GP (telling them you are participating in the study) and so that we can view you knee x-ray data.

Next, we will measure your height and weight and then ask you to change into shorts and a comfortable T-shirt. You will then complete a short questionnaire which will allow us to understand how much your knee arthritis interferes with your daily life. To begin with the researcher will place small electrodes over specific muscles on the front and back of your thigh, on your calf and on the front of your shin. Before each of these electrodes is positioned, the researcher will remove any excess hair with a disposable razor and then remove dead skin with an exfoliating cream. Once the electrodes are in place, the researcher will then position reflective markers on your legs, feet, arms and upper body using hypo-allergenic tape as shown in the picture to the right.

With the electrodes and markers in place the researcher is able to capture biomechanical data during movement. To begin with you will be asked to stand still whilst a standing trial is recorded. Following the standing trial, we will ask you to walk in a pair of normal shoes 5-10 times, over a distance of 6 metres, at your normal walking speed. We will then ask you to change into a pair of rocker shoes and to walk another 5-10 times, again at your normal walking speed. Finally, you will change into a pair of flexible shoes and again repeat the 5-10 walks. Note that the rocker shoes have a curved bottom and are designed to ‘rock’ the foot forward thereby making walking a little easier (see picture to the right).

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USE OF THE DATA

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CONTACT INFORMATION

If you require more information about the study, want to participate, or if you are already participating and want to withdraw, please contact

	Ali Algarni
Email:	A.S.S.Algarni@edu.salford.ac.uk
Phone :	07985502817
Address:	School of Health Sciences
	Brian Blatchford Building,
	University of Salford

Salford
Manchester M6 6PU

Thank you very much for taking time to read this document!

We appreciate your interest in this study and hope to welcome you at the School of Health Sciences, University of Salford.

Appendix 6: WOMAC questionnaires

Rocker shoe in Knee osteoarthritis

WOMAC OSTEOARTHRITIS INDEX VERSION LK3.0

INSTRUCTIONS TO PATIENTS

In Sections A, B and C questions will be asked in the following format and you should give your answers by putting an "X" in one of the boxes.

NOTE:

1. If you put your "X" in the left-hand box, i.e.

None	Mild	Moderate	Severe	Extreme
☒	☐	☐	☐	☐

then you are indicating that you have no pain.

2. If you put your "X" in the right-hand box, i.e.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☒

then you are indicating that your pain is extreme.

3. Please note:
   a) that the further to the right you place your "X" the more pain you are experiencing.

   b) that the further to the left you place your "X" the less pain you are experiencing.

   c) **please do not place your "X" outside the box.**

You will be asked to indicate on this type of scale the amount of pain, stiffness or disability you have experienced in the last 48 hours.

Remember the further you place your "X" to the right, the more pain, stiffness or disability you are indicating that you experienced. Finally, please note that you are to complete the questionnaire with respect to your study joint(s). You should think about your study joint(s) when answering the questionnaire, i.e., you should indicate the severity of your pain, stiffness and physical disability that you feel is caused by arthritis in your study joint(s). Your study joint(s) has been identified for you by your health care professional. If you are unsure which joint(s) is your study joint, please ask before completing the questionnaire.

Section A

INSTRUCTIONS TO PATIENTS

The following questions concern the amount of pain you have experienced due to arthritis in your study joint(s). For each situation please enter the amount of pain experienced in the last 48 hours. (Please mark your answers with an "X".)

QUESTION: How much pain do you have?

1. Walking on a flat surface.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

2. Going up or down stairs.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

3. At night while in bed.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

4. Sitting or lying.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

5. Standing upright.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PAIN1 ______

PAIN2 ______

PAIN3 ______

PAIN4 ______

PAIN5 ______

Section B

INSTRUCTIONS TO PATIENTS

The following questions concern the amount of joint stiffness (not pain) you have experienced in the last 48 hours in your study joint(s). Stiffness is a sensation of restriction or slowness in the ease with which you move your joints. (Please mark your answers with an "X".)

6. How severe is your stiffness after first wakening in the morning?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

7. How severe is your stiffness after sitting, lying or resting later in the day?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

STIFF6 ______

STIFF7 ______

Section C

INSTRUCTIONS TO PATIENTS

The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities, please indicate the degree of difficulty you have experienced in the last 48 hours due to arthritis in your study joint(s). (Please mark your answers with an "X".)

QUESTION: What degree of difficulty do you have?

8. Descending stairs.

None	Mild	Moderate	Severe	Extreme	
☐	☐	☐	☐	☐	PFTN8 ____

9. Ascending stairs.

None	Mild	Moderate	Severe	Extreme	
☐	☐	☐	☐	☐	PFTN9 ____

10. Rising from sitting.

None	Mild	Moderate	Severe	Extreme	
☐	☐	☐	☐	☐	PFTN10 ____

11. Standing.

None	Mild	Moderate	Severe	Extreme	
☐	☐	☐	☐	☐	PFTN11 ____

12. Bending to floor.

None	Mild	Moderate	Severe	Extreme	
☐	☐	☐	☐	☐	PFTN12 ____

13. Walking on flat.

None	Mild	Moderate	Severe	Extreme	
☐	☐	☐	☐	☐	PFTN13 ____

14. Getting in/out of car.

None	Mild	Moderate	Severe	Extreme	
☐	☐	☐	☐	☐	PFTN14 ____

15. Going shopping.

None	Mild	Moderate	Severe	Extreme	
☐	☐	☐	☐	☐	PFTN15 ____

16. Putting on socks/stockings.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PFTN16 ____

17. Rising from bed.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PFTN17 ____

18. Taking off socks/stockings.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PFTN18 ____

19. Lying in bed.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PFTN19 ____

20. Getting in/out of bath.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PFTN20 ____

21. Sitting.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PFTN21 ____

22. Getting on/off toilet.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PFTN22 ____

23. Heavy domestic duties.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PFTN23 ____

24. Light domestic duties.

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

PFTN24 ____

THANK YOU FOR COMPLETING THE QUESTIONNAIRE